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14. ABSTRACT The overall purpose of this research is to determine the effects of fatigue, load carriage, and history of physical activity on the mechanical loading of the tibia and risk of tibial stress fracture. There are two specific aims in this study. The first aim is to determine the influences of fatigue and load carriage on gait mechanics and tibial loading. Twenty participants were recruited to perform walking in both fatigued and loaded conditions. CT scans of lower legs were obtained. Kinematic and kinetic gait analysis, musculoskeletal modeling, and finite element analysis of the tibia loading were conducted. The second research aim is to determine the effects of history of physical activity on tibial mechanical loading during impact related activities. Forty participants were recruited to perform running, landing, cutting maneuvers, and walking with loads. CT scans of lower legs were obtained. Kinematic and kinetic movement analysis, musculoskeletal modeling, and finite element analysis of the tibia loading are conducted. Within this final report, information concerning adherence to work objectives, results, and reportable outcomes are presented. Overall, we have achieved work objectives of the research and tested all the hypotheses. The data analysis demonstrates encouraging results.				
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INTRODUCTION

Due to the enormity of the tibial stress fracture (TSF) problem in the military it is vitally important to develop a better understanding of how the conditions experienced by military personnel affect their risk for TSF. The overall purpose of this study was to determine how the factors of fatigue, load carriage, and history of physical activity affect the mechanical loading of the tibia and subsequent risk of tibial stress fracture. The research project had been conducted in two phases.

The first phase of this research focused on the effects of load carriage and fatigue on mechanical variables associated with increased risk of tibial stress fractures. Motion capture, ground reaction force, electromyographic, and computed tomography data were used to create participant specific musculoskeletal and finite element models. These models were used to determine the tibial strains and strain rates during both loaded and fatigued walking conditions. The immediate significance of this research is the determination of how different tasks and conditions affect the tibial strains produced during militarily relevant conditions. It is this mechanical loading and the strains it causes in the bone that ultimately leads to TSF. The results of this study can be used to evaluate the relative increases in bone strain and potential risk of TSF due to carrying additional loads, fatigue, and the combination of fatigue and load carriage. By understanding how load and fatigue influence tibial strains, the appropriate measures can be taken in the future to minimize the risks to recruits. This could be in form of altering load carriage, training programs, or a combination of the two.

The second phase of this research focused on the effects of previous physical activity history on the mechanical variables associated with increased risk of tibial stress fractures. Two groups of participants were recruited. Group 1 consisted of participants who had participated regularly in a minimum of 2 years of recreational basketball. Group 2 consisted of participants who had participated regularly in a minimum of 2 years of recreational running. Motion capture, ground reaction force, electromyographic, and computed tomography data were used to create participant specific musculoskeletal and finite element models. These models were used to determine the tibial strains and strain rates during 1) running, landing, and cutting maneuvers and 2) progressively loaded walking trials. The significance of this research was that it would provide information regarding how physical activity history affects tibial strains in conditions similar to those encountered in basic training. The results from this phase of the research can be used to determine if there are any benefits to engaging in multi-directional, irregular impact activities (such as those encountered in basketball) compared to running. This is relevant because one of the major issues yet to be resolved in stress fracture prevention is the determination of the type and frequency of exercise that is vigorous enough to stimulate bone growth and adaptation yet not severe enough as to lead to overuse injuries. This research will lead to advances in our understanding of how participation in different types of physical activity relates to the incidence of TSF. These data may then be used to develop more effective training programs and to better identify individuals who are at higher risk of sustaining stress fractures due to their history of physical activity.

The comprehensive scope of the proposed research will contribute significantly to the scarcity of published literature on in-vivo tibial strains and strain rates during activities consistent with what are required for military personnel. These data are needed in order to develop a better understanding of the dose-response relationship between loading conditions and bone health.

BODY

PHASE 1 PROJECT

There is evidence that load carriage and fatigue affect basic kinematic and kinetic gait parameters in a way that is consistent with increased mechanical demands on the tibia. However, the specific effects of load carriage and fatigue on tibial loading, specifically strain and strain rates, are not known. In order to determine these effects, a combination of load carriage and fatiguing protocols are examined using traditional gait analysis techniques in conjunction with new musculoskeletal modeling and finite element analysis (FEA) protocols.

The purpose of the first phase of this research is to determine the effects of load carriage and fatigue on mechanical variables associated with increased risk of tibial stress fractures. Currently, there is no clear consensus on how conditions such as carrying additional loads or fatigue influence the strains in the tibia.

Summary of Methods

In order to determine the effects of load carriage and fatigue on mechanical variables associated with increased risk of tibial stress fracture, traditional kinematic and kinetic analyses derived from motion capture and force plate data have been combined with a participant-specific finite element model of the tibia utilizing customized musculoskeletal modeling software and FEA (Figure 1).

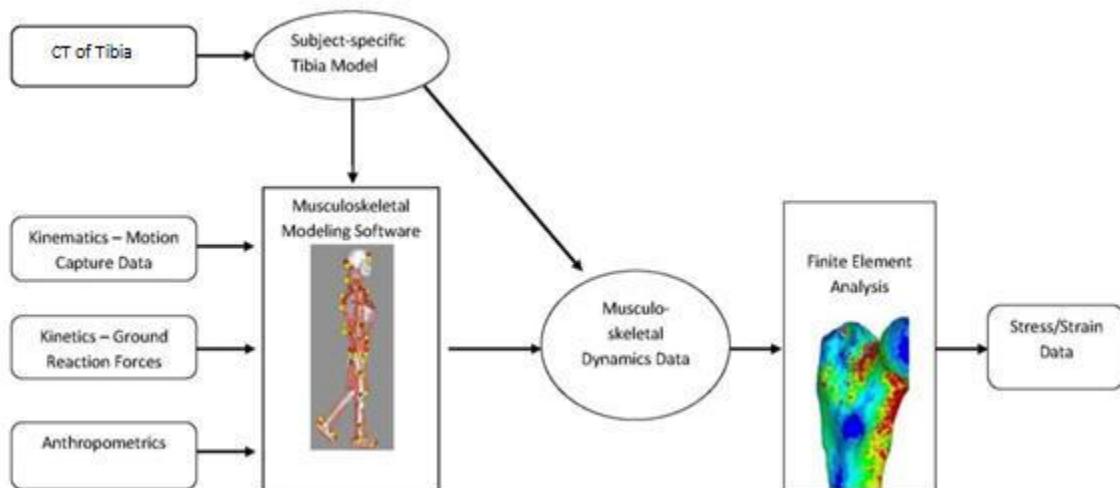


Figure 1. Diagram of project workflow

In order to test the Phase 1 hypotheses that 1) an increase in load carried will result in changes in gait mechanics and increased strains and strain rates in the tibia, and 2) fatigue will result in changes in gait mechanics and increased strains and strain rates in the tibia, the following protocols and procedures have been implemented.

Participants

In order to control for the confounding factors of hormonal status, body size, and strength, the study was conducted using a convenience sample of 20 male participants. Participants were recruited from the Ball

State University student population. Upon signing the written informed consent approved by the university's institutional review board for human subject participation in research and the HRPO of the U.S. Army Medical Research and Materiel Command, participants were scheduled and tested.

The participants completed three data collection sessions; 1) assessment of aerobic and muscular strength, 2) gait analysis during loaded/fatigued walking conditions, and 3) computed tomography (CT) imaging of their tibias.

Fitness Assessments

The participants completed an aerobic and muscular fitness assessment session which lasted approximately two hours.

First, an assessment of maximal oxygen consumption, or VO₂ peak, was made. Expired gases were collected with a ParvoMedics metabolic cart (ParvoMedics, Sandy, UT), which was calibrated prior to each testing session with standard gasses and a 3L syringe. A modified ramped version of the Bruce treadmill protocol was used [1]. The participants began the testing by walking at 1.7 mph on a 10% incline. Speed and grade were gradually increased until the subject voluntarily terminated the test due to fatigue. Resting heart rate and blood pressure were measured and monitored at regular intervals throughout the testing session. Next, the participants were given a 30 minute rest.

Following the rest period, the participants' lower body muscular strength was measured using a 1RM effort leg press similar to that described by Hoffman et al [2]. The participants self-selected an initial starting weight with which they could perform approximately 10 repetitions. After this initial set of 10 repetitions, the researcher in charge of the data collection increased the weight by 2.5 to 20kg, as described in the ACSM's Guidelines for Exercise Testing and Prescription [3]. Two minutes of rest was given between trials. The assessment was continued until the participants could no longer successfully complete a repetition. The weight of the last successful attempt was recorded as the participants' 1RM.

Load Carriage and Fatigue Protocol

The participants completed a set of walking trials under different loading and fatigue conditions during which standard motion capture data were collected.

Fifteen VICON M-series cameras (120 Hz) and VICON NEXUS 1.4 software (VICON Inc. Denver, CO) were used to collect three dimensional coordinates of reflective markers, ground reaction forces, and EMG of leg muscles. A modified standard plug-in gait marker set with cluster plates was used. A tandem forceplate instrumented treadmill (AMTI Inc., Watertown, MA) was used to set the walking speed at 1.67 m/s and allowed ground reaction forces to be collected at 2400 Hz. A 16-channel Delsys EMG system (Delsys Inc., Boston, MA) (2400 Hz) with bipolar single differential surface electrodes (41 mm x 20 mm electrode area, inter-electrode distance 1 cm) was used to monitor muscular activity of selected leg muscles.

Prior to data collection, basic participant anthropometric measurements were performed. These measurements included: height, weight, leg length, ankle width, knee width, and ASIS width.

Following the basic measurements and application of markers and electrodes, the participants warmed up in preparation for assessment of their maximal vertical jump height. The participants warmed up by walking at a self-selected pace on the treadmill for 5 minutes, followed by five practice jumps. The

participants' maximal vertical jump height was assessed using a Vertec, they performed three maximal vertical jumps and the highest of the three was recorded.

Following the baseline vertical jump test, the participants began the walking protocol as outlined below:

1. Walk for 5 minutes with no backpack
2. Walk for 5 minutes carrying a 32kg backpack*
3. Completion of Fatigue Protocol
4. Walk for 5 minutes fatigued carrying a 32kg backpack*
5. Walk for 5 minutes fatigued with no backpack

*This 32kg load is similar to the Army's suggested approach or marching load [4].

During each of these stages, the participants walked on a treadmill (0° incline) at a pace of 1.67m/s. The participants wore standard military boots (Altama™ Mil-Spec Desert 3 Layer boot), shorts, and t-shirt during the experiment. During the testing session, motion capture, ground reaction force, and EMG data were recorded for the last minute of each stage of the protocol.

The fatiguing protocol took place between the second and fourth stages of the walking protocols. Due to the potentially different effects of whole body fatigue vs. local muscle fatigue on factors relevant to tibial stress fracture [5, 6], a combination of activities were used.

The participants completed a circuit which included loaded stepping and heel raises. The stepping protocol was based on a Queens College Step Test procedure [7]. The participants performed the test while wearing a 16kg backpack. The step height remained at the standard 16 inches and the participants stepped up and down at a rate of 24 cycles per minute. One cycle was defined as step up with first leg, step up with contra-lateral leg, step down first leg, and step down with contra-lateral leg. A metronome was set at 4 times the cycle rate to correspond to leg movements, in this case 96 beats per minute. Participants performed this stepping sequence until they could no longer match the cadence of the metronome. At this time the participants completed 20 heel raises standing at the edge of a box. Following the heel raises the participants removed the backpack and completed a maximal effort vertical jump measured using the Vertec. This sequence was repeated until the participants' maximal vertical jump fell to or below 80% of its original value. Members of the research team provided verbal encouragement throughout the testing in order to elicit a maximal effort from the participants.

CT Imaging Protocol

The participants visited Ball Memorial Hospital (Muncie, IN) accompanied by a member of the research team. During this visit bilateral tibial CTs were recorded. This process took approximately one hour.

Axial plane scans were obtained using the following parameters: slice thickness 0.625mm, 15cm x 15cm field of view (FOV), kVp and mAs determined by machine algorithm (auto) based upon participant anthropometrics, each leg scanned separately from the distal femur through the calcaneous. Images were reconstructed in a 512x512 matrix using bone parameters derived from the literature. In addition to using the tibial CTs for modeling purposes, the images were used to calculate the tibial size and cross sectional area at the narrowest portion of the tibia, for consideration in subsequent analyses.

Modeling Methods

Bone geometries were obtained using computed tomography (CT). CT images were segmented and 3D geometry files generated in Materialise MIMICS 13.0 (Materialise, Leuven, Belgium).

The 3D surface geometries were used in MD MARC 2008 (MSC Software, Santa Anna, CA) to build hexmesh finite element (FE) models with generic linear isotropic material properties of elastic modulus 17GPa, density 1.9g/cm³, and Poisson's ratio of 0.3. Tibia models were then imported into a scaled LifeMOD model.

Once positioned, spatial coordinates of muscle model markers representing origins, insertions, and joint positions relative to the tibia were incorporated into the FE tibia. Boundary conditions were assigned as rotational and translational degrees of freedom of nodes representing ankle and knee joint centers and flexible bodies (modal neutral files) were generated for LifeMOD.

For a full description of LifeMOD modeling methods and use of flexible bodies, see Al Nazer [8]. Key differences between this study and that of Al Nazer et al included more muscle actuators on the leg with the flexible tibia, participant specific segment scaling based on joint center calculations, ground reaction forces, and participant specific tibial geometries generated from CT scans.

A lower body model was built using participant sex, mass, and height as scaling inputs in the GeBOD database [9]. Segments, joint locations, and orientations were scaled using joint center data from Visual 3D. Experimental kinematic data were used to perform an inverse kinematic analysis. Results of this analysis were used to "train" the muscle (right leg) and joint (left leg) PID controllers used to actuate the model in a forward dynamics (FD) analysis. These controllers were "trained" by using the kinematics of the inverse kinematics run as the targets for the forces produced by the actuators. If these targets were not met, force production was modulated to better produce the kinematics. Flexible tibias were then imported into LifeMOD. A FD analysis was performed with the addition of ground reaction forces applied to the feet and motion capture kinematics disabled. Maximum principal, minimum principal, and maximum shear strain values were then calculated using the Durability plug-in for MD ADAMS/View for the nodes of the geometric middle third of the tibial shaft (MSC Software, Santa Ana, CA).

Statistical Analyses

The statistical analysis was divided into two parts, one focusing on traditional gait mechanics and the other on the tibial strain profiles. While these analyses both address the general research question regarding the impact of load and fatigue on risk factors associated with tibial stress fracture, the differences in the research designs used to gather the data require the use of somewhat different statistical modeling.

Traditional Gait and Loading Mechanics

In order to test whether load, fatigue or the combination of these factors results in changes in gait mechanics; the following dependent (outcome) variables were calculated and analyzed:

- Peak vertical ground reaction forces
- Vertical ground reaction force loading rate
- Stride frequency

- Double support time

The target independent variables were loading condition and fatigue, which were completely crossed. In addition to these primary independent variables, a number of other variables (referred to collectively as covariates) were included in the analysis in order to control statistically for factors that had previously been shown to be associated with the risk of tibial stress fracture. These covariates included body mass, leg length, tibial size (cross sectional area (CSA) at the narrowest portion of the tibia) and fitness level as measured by 1RM leg press and VO₂ peak.

In order to assess the impact of fatigue and loading condition on the outcome variables, a repeated-measures Analysis of Covariance (ANCOVA) was used. ANCOVA was appropriate for this statistical analysis because it accounted for the continuous independent variables that were collected from each individual and used to control for variables known to be related to tibial stress fractures. Because the participants were measured under two different load conditions, and two levels of fatigue, the repeated measures aspect of the analysis allowed for a direct comparison of dependent variable means across load conditions and fatigue levels, and allowed for an assessment of the interaction of these two variables.

Tibial Strain Profile

In order to determine whether load, fatigue or the combination of these factors results in changes of the tibial strain profile, the following dependent (outcome) variables were calculated and analyzed:

- Peak strain in the middle third of the tibia
- Peak strain rate in the middle third of the tibia

The independent variables and covariates of interest for this analysis were the same as for traditional gait mechanics, with the primary focus being on fatigue and loading condition and their interaction. The actual modeling of the dependent variables was done using a repeated-measures ANCOVA, as described above. As with the traditional gait mechanics, this analysis provided significance tests for the primary independent variables while controlling for the covariates and the repeated measurements on the same individuals.

Combining the results from these two analyses enabled the researchers to determine whether or not the data collected in this phase of the study support in full or in part the hypotheses that 1) an increase in load carried will result in changes in gait mechanics and increased strains and strain rates in the tibia and 2) fatigue will result in changes in gait mechanics and increased strains and strain rates in the tibia.

PHASE 2 PROJECT

There is evidence that an individual's past physical activity influences his or her risk of sustaining a tibial stress fracture. This evidence is based primarily on epidemiologic research on the rates of injury in different sub-populations. The mechanisms that may explain these results have not been adequately examined. In order to assess the effect of previous physical activity, two groups of individuals, basketball players, and runners completed a loaded walking protocol along with a series of other high impact activities such as running, drop-jumping, and cutting. These tasks were analyzed using traditional gait analysis techniques in conjunction with subject-specific musculoskeletal modeling and FEA protocols in order to determine if differences in tibial strains and strain rates exist between the two groups.

The purpose of the second phase of this research was to determine the effects of physical activity history on mechanical variables associated with increased risk of tibial stress fractures.

Summary of Methods

In order to determine the effects of physical activity history on mechanical variables associated with increased risk of tibial stress fracture, traditional mechanical analyses derived from motion capture and force plate data had been combined with a subject-specific finite element model of the tibia utilizing customized musculoskeletal modeling software and FEA (Figure 1).

In order to test the following hypotheses: 1) regular participation in sports that involve irregular, high impact cutting and landing maneuvers for minimum of 2 years results in bone adaptations that lead to lower strains in the tibia during loaded walking tasks. 2) high impact, irregular sporting maneuvers such as cutting and landing that are performed during playing basketball, produce different tibial strains and strain rate patterns than those produced during running. The following protocols and procedures have been implemented.

Participants

In order to control for the confounding factors of hormonal status, body size, and strength, the study was conducted using a convenience sample of two groups of 20 male participants each, for a total of 40 participants. Participants were recruited from the Ball State University student population. Group 1 consisted of participants who had participated regularly in a minimum of 2 years of recreational basketball. Group 2 consisted of participants who had participated regularly in a minimum of 2 years of recreational running. Upon signing the written informed consent approved by the university's institutional review board for human subject participation in research and the HRPO of the U.S. Army Medical Research and Materiel Command, the participants were scheduled and tested.

The participants completed three data collection sessions; 1) assessment of aerobic and muscular strength, 2) motion capture, force plate, and electromyographic (EMG) data collection while performing drop-jumping, cutting, running, and loaded walking tasks, and 3) computed tomography imaging of their tibias.

Fitness Assessments

The participants completed an aerobic and muscular fitness assessment session which lasted about two hours. The fitness assessment protocol was the same as Phase 1 research and was described in phase 1 research protocol.

Motion Capture Protocol

Fourteen VICON M-series and F-series cameras (240 Hz) and VICON Workstation 5.0 software (VICON Inc. Denver, CO) were used to collect three dimensional coordinates of reflective markers, ground reaction forces, and EMG of leg muscles. A modified standard plug-in gait marker set with cluster plates was used. A tandem forceplate instrumented treadmill (AMTI Inc., Watertown, MA) was used to set the walking speed at 6 km/h and running speed at 12 km/h and allowed ground reaction

forces to be collected at 2400 Hz. A 16-channel Delsys EMG system (Delsys Inc., Boston, MA) (2400 Hz) with bipolar single differential surface electrodes (41 mm x 20 mm electrode area, inter-electrode distance 1 cm) was used to monitor muscular activity of selected leg muscles.

Prior to data collection, basic participant anthropometric measurements were performed. These measurements included: height, weight, leg length, ankle width, knee width, and ASIS width.

Following the basic measurements and application of markers and electrodes, the participants warmed up in preparation for assessment of their maximal vertical jump height. The participants warmed up by walking at a self-selected pace on the treadmill for 5 minutes, followed by five practice jumps. The participants' maximal vertical jump height was assessed using a Vertec, they performed three maximal vertical jumps and the highest of the three was recorded.

Following the baseline vertical jump test, the participants performed the following movement tasks while motion capture, force plate, and EMG data were collected:

Drop-jump: participants performed a short series of drop-jumps to mimic jump and landing task consistent with those performed in basketball. The jump height was based on 80% of the participant's max vertical jump height. Participants performed 10 drop-jumps onto two force plates (one foot landing on each plate). Thirty seconds rest was given between jumps.

Cutting Maneuver: participants performed a cutting maneuver described by Sigward et al. [10]. Participants were instructed to run for 5m and then performed 45-degree cutting maneuvers to the left for five trials and to the right for five trials. A one-minute rest was given between cuttings.

Unloaded Running: participants performed unloaded running at 12 km/h on a force instrumented treadmill (AMTI, Watertown, MA). The participants started out walking at 6 km/h for 2 minutes, and then increased speed 1 km/h each minute until the target speed of 12 km/h was reached. The participants maintained the speed for 5 minutes. Three 10 second trials were collected for analysis.

Loaded Walking: participants walked at 6km/h on an instrumented treadmill (AMTI, Watertown, MA) with the following loads carried: 0kg, 15kg, 25kg, and 35kg. Participants walked for 5 minutes during each loaded walking condition. Three 10 second trials were collected during each walking task. A five-minute rest was given between walking conditions.

CT Imaging Protocol

The participants visited Ball Memorial Hospital (Muncie, IN) accompanied by a member of the research team. During this visit bilateral tibial CTs were recorded. This process took approximately one hour. The CT imaging protocol was the same as Phase 1 research and was described in phase 1 research protocol.

Modeling Methods

Tibia geometries were obtained using computed tomography (CT). A custom build calibration phantom was used to establish a relationship between the CT Hounsfield units and tibial bone density. A relationship between bone density and Young's modulus was established based on the literature [11]. CT images were segmented and 3D geometry files were generated in MIMICS 14.0 (Materialise, Leuven, Belgium).

The 3D surface geometries of tibias were used in MARC 2012 (MSC Software, Santa Anna, CA) to build hexmesh finite element (FE) models. Material properties including bone density, Young's modulus, and Poisson's ratio were assigned to each of the individual elements of the tibial FE model. The FE tibia models were then imported into the subject-specific musculo-skeletal models in LifeMOD.

Once positioned, spatial coordinates of muscle model markers representing origins, insertions, and joint positions relative to the tibia were incorporated into the FE tibia. Boundary conditions were assigned as rotational and translational degrees of freedom of nodes representing ankle and knee joint centers and flexible bodies (modal neutral files) were generated in MARC 2012 for LifeMOD simulation.

A lower body model was built using participant sex, mass, and height as scaling inputs in the GeBOD database [9]. Segments, joint locations, and orientations were scaled using joint center data from Visual3D. Experimental kinematic data were used to perform an inverse kinematic analysis. Results of this analysis were used to "train" the muscles of both legs, which were used to actuate the model in a forward dynamics (FD) analysis. These muscles were "trained" by using the kinematics of the inverse kinematics run as the targets for the forces produced by the actuators. If these targets were not met, force production would be modulated to better produce the kinematics. Flexible tibias were then imported into LifeMOD. A FD analysis was performed with the addition of ground reaction forces applied to the feet and motion capture kinematics disabled. Maximum principal, minimum principal, and maximum shear strain values were calculated using the Durability plug-in for ADAMS/View 2012 for the surface nodes of the tibial shaft (MSC Software, Santa Ana, CA).

Statistical Analyses

The statistical analyses for phase 2 project were carried out using two separate models. The first of these analyses focused on the impact of increasing load carried and type of athletic training on the strain profiles across the tibia, while the second part focused on the magnitude and location of peak strain and strain rates across the tibia as a function of type of athletic activity. Following is a description of each statistical analysis used in phase 2 project.

Tibial Strain and Strain Rate by Loading Condition and Previous Physical Activity

In order to test the hypothesis that regular participation in basketball for a minimum of 2 years results in bone adaptation that lead to lower strains in the tibia during loaded walking tasks, the following approach was used.

The dependent variables of interest were peak strain and strain rate in each third of the tibial shaft. The independent variables were loading condition (walking with 0, 15, 25, 35 kg loads), and physical activity history (basketball or running). In addition, location on the tibia (proximal third, middle third, or distal third) was included in the analysis.

In order to ascertain the impact of load and physical activity history on peak strain and strain rate, repeated-measures ANOVAs were performed. There were two within-subject factors, load condition and location on the tibia, and one between-subject factor, type of previous physical activity. This analysis included tests of the three main effects, thus allowing for an assessment of whether there were significant differences in peak strains and strain rates across loadings, type of physical activity history and location on the tibia.

Peak Tibial Strain and Strain Rate by Type of Athletic Activity

In order to test the hypothesis that high impact, irregular sporting maneuvers such as cutting and landing, which were consistent with those performed in basketball, produce different tibial strain and strain rate patterns than those produced during running, repeated-measures ANOVA tests were performed.

The dependent variables were peak strains and strain rates. The independent variables were type of athletic activity, physical activity history, and location on the tibia. The repeated-measures ANOVAs had two within-subject factors (type of athletic activity and location on the tibia) and one between-subject factor (previous physical activity). This ANOVA test provided hypothesis tests for the equality of mean values across type of athletic activity, type of physical activity history and location on the tibia for each dependent variable.

KEY RESEARCH ACCOMPLISHMENTS

- The abstract titled “*An integrated modeling method for tibia strain analysis*” was presented at the Northwest American Society of Biomechanics Symposium in May 2010 (Appendix A)
- The abstract titled “*Muscular fatigue increases ground reaction loading rate during walking*” was presented at the 57th Annual Meeting of American College of Sports Medicine in June 2010. The abstract was published in *Medicine & Science in Sports & Exercise*. V. 42, No. 5 Supplement, S192 (Appendix B)
- The abstract titled “*Load carriage increases mechanical loading rates during walking*” was presented at the 34th Annual Meeting of American Society of Biomechanics in August 2010 (Appendix C)
- The abstract titled “*An integrated modeling method for bone strain analysis*” was presented at the 34th Annual Meeting of American Society of Biomechanics in August 2010 (Appendix D)
- The abstract titled “*Effects of load carriage and muscular fatigue on ground reaction loading rate during walking*” was presented at the 58th Annual Meeting of American College of Sports Medicine in June 2011. The abstract was published in *Medicine & Science in Sports & Exercise*. V. 43, No. 5 Supplement, S21-22 (Appendix E)
- The abstract titled “*The effect of height on tibial strain while performing drop landings*” was submitted and accepted for presentation at the 58th Annual Meeting of American College of Sports Medicine in June 2011. The abstract was published in *Medicine & Science in Sports & Exercise*. V. 43, No. 5 Supplement, S639 (Appendix F)
- The abstract titled “*Influences of load carriage and fatigue on lower-extremity kinetics during walking*” was presented at the 35th Annual Meeting of American Society of Biomechanics in August 2011 (Appendix G)
- The abstract titled “*A time-efficient method for analyzing bone strain with large subject pools*” was presented at the 35th Annual Meeting of American Society of Biomechanics in August 2011 (Appendix H)
- The manuscript titled “*Influence of fatigue and load carriage on mechanical loading during walking*” was published in the *Journal of Military Medicine* in January 2012 (Appendix I)
- The abstract titled “*Influence of physical activity history on ground reaction force during walking*” was presented at the 59th Annual Meeting of American College of Sports Medicine in May 2012. The abstract was published in *Medicine & Science in Sports & Exercise*. V. 44, No. 5 Supplement, S282-283 (Appendix J)
- The abstract titled “*The effects of load carriage and fatigue on frontal plane knee mechanics during walking*” was presented at the 36th Annual Meeting of American Society of Biomechanics in August 2012 (Appendix K)

- The abstract titled “*the influence of physical activity history on ground reaction force during running*” was presented at the 60th Annual Meeting of American College of Sports Medicine in May 2013. The abstract was published in *Medicine & Science in Sports & Exercise*. V. 45, No. 5 Supplement, S502 (Appendix L)
- The manuscript titled “*The effects of load carriage and muscle fatigue on lower extremity joint mechanics*” was published in the *Research Quarterly of Exercise and Sport* in September 2013 (Appendix M)
- The abstract titled “*the effects of the type of activity on tibial strain characteristics*” was presented at the 37th Annual Meeting of American Society of Biomechanics in September 2013. (Appendix N)

REPORTABLE OUTCOMES for THE PHASE ONE PROJECT

The following sections outline the results from the first phase of this project. These data along with subsequent analyses will form the basis for forthcoming presentations and manuscripts documenting this research.

Participant Characteristics:

Twenty college-age male participants were recruited to participate in the first phase of this study. Technical difficulties were encountered with data from two of the participants, whose data were excluded from the analyses. Therefore the final data analysis was performed on 18 participants; their characteristics are presented in Table 1.

Table 1. Subject demographic information

	Mean (SD)
Age (yrs)	20.8 (1.8)
Height (cm)	180.8 (4.5)
Mass (kg)	77.6 (9.6)
Minimum Tibia CSA (mm ²)	364 (49)
Leg Press 1RM (lbs)	529 (169)
VO ₂ Peak (mL/kg/min)	51 (5)

Gait Kinematics and Kinetics:

Gait kinematics and kinetics of the four walking tasks performed are presented in table 2 and table 3. Table 2 shows the means and standard deviations (SDs) of stride frequency, stride length, and double support time. Table 3 shows the means and SDs of the peak vertical and braking ground reaction forces and loading rates during weight acceptance of walking. Two-way repeated-measures ANOVAs were used to determine the effects of load carriage and muscle fatigue on the selected gait kinematics and kinetics variables.

Table 2. Means (SDs) of spatio-temporal parameters during walking

Variables\Conditions	UU	UF	LU	LF
Stride Frequency (strides/min)	58.6 (2.7)	59.7 (4.0)	60.5 (2.4)	63.4 (5.1)
Stride Length (m)	1.71 (0.07)	1.68 (0.11)	1.66 (0.06)	1.59 (0.12)
Double Support (% gait cycle)	22.1 (1.4)	23.0 (1.5)	27.9 (2.1)	29.5 (2.9)

Conditions: UU = unloaded unfatigued, UF = unloaded fatigued, LU = loaded unfatigued, LF = loaded fatigued.

Table 3. Means (SDs) of the peak GRF and loading rates during walking

Variables\Conditions	UU	UF	LU	LF
Peak VGRF (BW)	1.27 (0.06)	1.35 (0.11)	1.92 (0.18)	1.99 (0.19)
Peak BGRF (BW)	0.23 (0.03)	0.24 (0.03)	0.35 (0.06)	0.34 (0.06)
Peak VGRLR (BW/s)	16.81 (3.40)	21.75 (7.92)	35.29 (12.07)	37.58 (11.92)
Peak BGRLR (BW/s)	7.98 (1.61)	9.78 (2.06)	14.81 (5.89)	15.44 (4.36)

Conditions: UU = unloaded unfatigued, UF = unloaded fatigued, LU = loaded unfatigued, LF = loaded fatigued.

VGRF = vertical ground reaction force, BGRF = braking ground reaction force, VGRLR = vertical ground reaction loading rate, BGRLR = braking ground reaction loading rate.

There was no interaction between the load carriage and fatigued conditions for all the gait kinematics and kinetics ($p > 0.05$). Load carriage had a significant effect on gait spatial-temporal parameters ($p < 0.01$) and kinetics ($p < 0.001$). Muscle fatigue also had a significant effect on gait spatial-temporal parameters ($p < 0.001$) and kinetics ($p < 0.001$).

Specifically, load carriage led to significant increases of stride frequency ($p = 0.007$) and double support time ($p = 0.004$) and a significant decrease of stride length ($p = 0.006$). Muscle fatigue resulted in significant increases of stride frequency ($p = 0.001$) and double support time ($p < 0.001$) and a significant decrease of stride length ($p = 0.001$).

Both load carriage and muscle fatigue resulted in alterations of ground reaction force variables. The load carriage led to significant increases of peak vertical and braking ground reaction forces ($p < 0.001$). Muscle fatigue led to a significant increase of peak vertical ground reaction force ($p < 0.001$). Furthermore, the load carriage led to significant increases of the peak vertical and braking ground reaction loading rates ($p < 0.001$). With a significant effect on peak vertical ground reaction loading rate ($p = 0.003$) and a near significant effect on peak braking ground reaction loading rate ($p = 0.084$), the muscle fatigue resulted in pronounced increases of ground reaction loading rates.

Lower-extremity joint kinematics and kinetics

Lower-extremity joint kinematics and kinetics at weight acceptance of the four walking tasks are presented in Table 4 and Table 5. Table 4 shows the means and SDs of the pelvis, hip, knee, and ankle angles at heel contact and stance of walking. Table 5 shows the means and SDs of the hip, knee, and ankle joint moment and power during weight acceptance of walking. Two-way repeated-measures ANOVAs were used to determine the effects of load carriage and muscle fatigue on the selected joint kinematic and kinetics during walking.

Table 4. Means (SDs) of the lower-extremity kinematics during walking

Variables\Conditions	UU	UF	LU	LF
Pelvis tilt at heel contact (deg)	8.8 (5.9)	11.0 (8.6)	20.1(5.0)	22.9 (8.6)
Hip flexion at heel contact (deg)	32.1 (4.3)	28.2 (10.4)	45.4 (5.2)	40.6 (10.9)
Knee flexion at heel contact (deg)	-2.5 (3.1)	-1.1 (4.5)	3.9 (3.2)	4.7 (4.9)
Maximum knee flexion at stance (deg)	19.0 (2.8)	20.7 (4.4)	24.6 (4.5)	25.0 (5.3)
Ankle dorsi-flexion at heel contact (deg)	7.7 (1.9)	5.3 (4.9)	7.3 (2.7)	5.6 (3.6)

Conditions UU = unloaded unfatigued, UF = unloaded fatigued, LU = loaded unfatigued, LF = loaded fatigued.

Table 5. Means (SDs) of the lower-extremity joint moments during walking

Variables\Conditions	UU	UF	LU	LF
Hip extensor moment (Nm/kg)	1.54 (0.41)	1.85 (0.48)	2.26 (0.42)	2.38 (0.42)
Knee extensor moment (Nm/kg)	0.88 (0.20)	0.90 (0.25)	1.61 (0.37)	1.63 (0.42)
Ankle dorsi-flexion moment (Nm/kg)	-0.44 (0.12)	-0.37 (0.10)	-0.45 (0.11)	-0.43 (0.11)
Hip joint power production (W/kg)	1.03 (0.36)	1.47 (0.49)	1.69 (0.61)	1.97 (0.64)
Knee joint power absorption (W/kg)	-1.34 (0.37)	-1.39 (0.54)	-2.65 (1.40)	-2.76 (1.06)
Ankle joint power absorption (W/kg)	-1.04 (0.27)	-0.82 (0.32)	-1.12 (0.30)	-1.01 (0.35)

Conditions UU = unloaded unfatigued, UF = unloaded fatigued, LU = loaded unfatigued, LF = loaded fatigued.

There was no interaction between the effects of load carriage and muscle fatigue for lower-extremity joint kinematics and kinetics ($p > 0.05$). Load carriage had a significant effect on lower-extremity joint kinematics and kinetics ($p < 0.001$). Muscle fatigue had a significant effect on lower-extremity joint kinematics and kinetics ($p < 0.05$).

Both load carriage and muscle fatigue led to pronounced alteration of lower-extremity joint kinematics. Load carriage resulted in significant increases of pelvis anterior tilt ($p < 0.001$), hip flexion ($p < 0.001$) and knee flexion ($p < 0.001$) at heel contact, and maximum knee flexion ($p < 0.001$) during stance. Muscle fatigue led to significant decrease of ankle-dorsi flexion at heel contact ($p = 0.028$).

Both load carriage and muscle fatigue led to pronounced alterations of lower-extremity joint kinetics. Load carriage led to significant increases of hip and knee extensor moments ($p < 0.001$) and a near significant increase of ankle dorsi-flexor moment ($p = 0.08$) during weight acceptance. Also, greater hip joint power production ($p < 0.001$), and knee and ankle joint power absorption ($p < 0.001$) were observed at weight acceptance of loaded walking. Muscle fatigue led to a significant increase of hip extensor moment ($p = 0.001$) and a significant decrease of ankle dorsi-flexor moment at weight acceptance. In addition, greater hip joint power production ($p = 0.001$) and lesser ankle joint power absorption ($p = 0.007$) at weight acceptance were observed during fatigued walking.

Strain and Strain Rate:

The peak strain and strain rates from the bone shaft of the tibia during unloaded unfatigued walking in the present study are presented along with *in-vivo* and simulated strains that have been reported by other researchers (table 6). The strains and strain rates are in reasonable agreement with the previously reported values. Differences in values between the current study and the previous studies may be due to the locations on tibia where the strain data were collected. Strain data reported in previous studies were from the antero-medial aspect of the tibia shaft [8, 12-15], while the current study uses average strains of the tibia shaft.

Table 6. Tibial strains and strain rates reported in previous studies and current study

	Strain Magnitude (Microstrain)			Strain Rate (Microstrain/s)		
	Max Principal	Min Principal	Max Shear	Max Principal	Min Principal	Max Shear
Lanyon et al. [12]	395	-434	829	Not reported	-4000	Not reported
Burr et al. [13]	437	-544	871	11006	-7183	16162
Milgrom et al. [14]	840	-454	1183	3955	-3306	10303
Milgrom el al. [15]	394	-672	Not reported	4683	-3820	Not reported
Al Nazer et al.[8]	305	-645	948	4000	-7000	10000
Present Simulation (Bone Shaft)	277	-409	655	4072	-5571	8986

A mixed model ANCOVA was used to determine if there were any differences in strains and strain rates between conditions. Significance was set at $\alpha = 0.05$. Height, body mass, age, minimum tibia cross sectional area (CSA), leg press 1RM, VO_2 peak were used as covariates.

Strain

Table 7. Means (SEs) of tibial strains during walking

Variables\Conditions	UU	UF	LU	LF
Max Principal Strain (μ s)	277.37 (90.81)	257.26 (90.81)	626.06 (90.81)	340.38 (90.81)
Min Principal Strain (μ s)	-409.26 (222.54)	-386.94 (222.54)	-1220.31 (222.54)	-519.29 (222.54)
Max Shear Strain (μ s)	654.95 (312.91)	612.58 (312.91)	1809.08 (312.91)	824.48 (312.91)

Conditions UU = unloaded unfatigued, UF = unloaded fatigued, LU = loaded unfatigued, LF = loaded fatigued.

Table 7 shows the means and standard errors (SEs) of the maximum and minimum principal strains and shear strains of the tibia for the four walking conditions. Load carriage had a significant effect on maximum principal strain ($p < 0.0001$), minimum principal strain ($p < 0.0001$), and maximum shear strain ($p < 0.0001$). Specifically, load carriage led to significant increases of maximum principal strain, minimum principal strain, and maximum shear strain during walking ($p < 0.0001$).

Muscle fatigue had a significant effect on maximum principal strain ($p < 0.0001$), minimum principal strain ($p < 0.0001$), and maximum shear strain ($p < 0.0001$). Specifically, muscle fatigue led to significant decreases of maximum principal strain, minimum principal strain, and maximum shear strain during walking ($p < 0.0001$).

None of the participant characteristics used as covariates (height, mass, age, leg press max, VO₂peak, or minimum tibial CSA) were significantly related to strain variables ($p > 0.05$).

Strain Rate

Table 8. Means (SEs) of tibial strain rates during walking

Variables\Conditions	UU	UF	LU	LF
Max Principal Strain Rate (μ s/s)	4071.62 (382.14)	3726.17 (382.14)	4455.86 (382.14)	3914.38 (382.14)
Min Principal Strain Rate (μ s/s)	-5570.64 (719.70)	-5208.27 (719.70)	-7179.86 (719.70)	-5793.44 (719.70)
Max Shear Strain Rate (μ s/s)	8985.66 (1057.61)	8248.04 (1057.61)	10859 (1057.61)	8943.08 (1057.61)

Conditions UU = unloaded unfatigued, UF = unloaded fatigued, LU = loaded unfatigued, LF = loaded fatigued.

Table 8 shows the means (SEs) of the maximum and minimum principal strain rates and shear strain rates of the tibia for the four walking conditions. Load carriage had a significant effect on maximum principal strain rate ($p < 0.0001$), minimum principal strain rate ($p < 0.0001$), and maximum shear strain rate ($p < 0.0001$). Specifically, load carriage led to significant increases of maximum principal strain rate, minimum principal strain rate, and maximum shear strain rate during walking ($p < 0.0001$).

Muscle fatigue had a significant effect on maximum principal strain rate ($p < 0.0001$), minimum principal strain rate ($p < 0.0001$), and maximum shear strain rate ($p < 0.0001$). Specifically, muscle fatigue led to significant decreases of maximum principal strain rate, minimum principal strain rate, and maximum shear strain rate during walking ($p < 0.0001$).

None of the participant characteristics used as covariates (height, mass, age, leg press max, VO₂ peak, or minimum tibial CSA) were significantly related to strain rates ($p > 0.05$).

SUMMARY and DISCUSSION:

We had hypothesized that load carriage and muscle fatigue would alter basic gait spatial-temporal parameters and kinetics. Results from this research support this hypothesis. Specifically, load carriage and muscle fatigue result in significant increases of stride frequency and double support time and a significant decrease of stride length. Furthermore, both load carriage and muscle fatigue lead to significant increases of the vertical ground reaction force and loading rate.

It was also hypothesized that load carriage and muscle fatigue would alter lower-extremity joint mechanics. This hypothesis was supported. Specifically, load carriage leads to significant increases of pelvis tilt, hip and knee flexion at heel contact and maximum knee flexion during stance. Muscle fatigue leads to a significant decrease of ankle dorsi-flexion at heel contact. Furthermore, load carriage leads to significant increases of hip and knee joint moment and hip, knee, and ankle joint powers. Muscle fatigue also leads to significant increases of hip joint moment and power and significant decreases of ankle joint moment and power.

Thus, when walking with load carried, lower-extremities experience significant increases of external impact forces (ground reaction forces) and loading rates at every foot strike. Moreover, the pronounced increases of lower-extremity joint moments and powers at weight acceptance reflect large increases of leg muscle forces. Therefore, lower-extremity skeletal structures (e.g. tibias) are loaded with both increased ground impact forces and increased leg muscles forces at stance of walking. In addition, the increased stride frequency and double support time associated with load carriage lead to increases of number of loading cycle and loading duration during walking.

Similarly, when walking in a fatigued state, lower-extremities experience significant increases of external impact force (vertical ground reaction force) and loading rate at each foot strike. The large increases of hip joint moment and power reflect increased effort of hip extensors to stabilize the pelvis. The significant decreases of ankle joint moment and power indicate reduced ability of the ankle dorsi-flexor to control the ankle joint at weight acceptance. Thus, lower legs experience increased ground impact force and loading rate and altered leg muscle loads at stance in a fatigued state. In addition, the increased stride frequency and double support time associated with muscle fatigue expose the lower-extremities to increased number of loading cycle and loading duration.

The combined effect of muscle fatigue and load carriage further exposes the lower-extremity skeletal structures to significant increases of external impact loading and loading rate and pronounced alterations of leg muscle loads. This pronounced alteration in mechanical loading pattern is coupled with increased number of loading cycle and loading duration during walking.

Furthermore, it was hypothesized that the changes in gait mechanics due to load carriage would result in increases in strains and strain rates in the tibia. Results from this study support this hypothesis. We found that the load carriage results in larger tibial strains and strain rates than walking with no loads. Repetitive high bone strains and strain rates were considered to be the etiology of stress fracture [13, 16]. Thus, results from this study provide evidence that load carriage could increase the risk of developing tibial stress fracture.

We also hypothesized that the muscle fatigue would result in increases in strains and strain rates in the tibia. This hypothesis was not supported. We found the muscle fatigue led to pronounced reductions in tibial strains and strain rates during walking. Results from this study were different from a previous *in vivo* study performed by Milgrom et al [15], which reported that fatigue resulted in pronounced increases of tensile strain, tensile and compressive strain rates during walking. However, the compressive strain was found to be significantly reduced during fatigued walk [15]. The discrepancies between the current study and Milgrom's study may be due to the following factors. Firstly, in Milgrom's study, the vertical ground reaction force (GRF) was found to be lower during fatigued walk than during normal walk. However, the current study found that the muscle fatigue leads to a significant increase in the vertical GRF. Secondly, in Milgrom's study, the walking speed was not controlled and not reported for the pre and post fatigue tests. In the current study, the walking speed was controlled at 1.67m/s for all the walking conditions. Thirdly, Milgrom's study did not present gait kinematic data. It is not known whether subjects participated in Milgrom's study walked differently from subjects recruited in this study. Thus, the possible differences in gait kinematics and ground reaction forces observed in the current study and Milgrom's study may contribute to the different findings in strains and strain rates.

In conclusion, load carriage and muscle fatigue result in alterations in gait mechanics signified by increases in stride frequency, double support time, vertical ground reaction force and loading rate. Load carriage leads to pronounced increases in tibial strains and strain rates. Muscle fatigue leads to pronounced decreases in tibial strains and strain rates.

REPORTABLE OUTCOMES for THE PHASE TWO PROJECT

The following sections outline the results from the second phase of this project. These data along with subsequent analyses will form the basis for forthcoming presentations and manuscripts documenting this research.

Participant Characteristics:

Forty college-age male participants were recruited in this study. Participants were divided into two experimental groups (runners and basketball players). The runner group consisted of twenty recreational runners with a minimum of two years of regular running experience. The basketball group consisted of twenty recreational basketball players with a minimum of two years of regular basketball playing experience. Participants' characteristics are presented in Table 9.

Table 9. Subject demographic information

	Runners	Basketball Players
Age (yrs)	20.7(2.4)	20.6(1.7)
Height (cm)	180.1(6.3)	180.8(8.3)
Mass (kg)	75.7(9.6)	80.1(11.5)
Minimum Tibia CSA (mm ²)	352(49)	369(49)
Leg Press 1RM (lbs)	492(62)	550(127)
VO ₂ Peak (mL/kg/min)	58(7)	50(5)

Mean (+/- standard deviation) for basic participant demographics and results from fitness assessments.

Kinematics and Kinetics of Loaded Walking Tasks:

Selected gait spatio-temporal parameters and kinetics of the four loaded walking tasks are presented in table 10, which shows the means and SDs of stride frequency, stride length, peak vertical ground reaction force and loading rate. Two-way repeated-measures ANOVAs were used to determine the effects of physical activity history and levels of loaded walking on the selected gait kinematics and kinetics variables.

Table 10. Means (SDs) of spatio-temporal parameters during loaded walking

Variables\Conditions	WK00	WK15	WK25	WK35
<u>Runners</u>				
Stride length (m)	1.70 (0.06)	1.69 (0.06)	1.67 (0.07)	1.63 (0.08)
Stride frequency (strides/min)	59 (2)	59 (2)	60 (2)	62 (3)
Peak VGRF (BW)	1.29 (0.07)	1.58 (0.10)	1.78 (0.15)	2.01 (0.19)
Peak VGRLR (BW/s)	18.19 (3.73)	21.45 (4.64)	26.53 (7.73)	32.30 (8.24)
<u>Basketball Players</u>				
Stride length (m)	1.69 (0.08)	1.68 (0.08)	1.66 (0.09)	1.62 (0.08)
Stride frequency (strides/min)	59 (3)	60 (3)	61 (3)	62 (3)
Peak VGRF (BW)	1.30 (0.05)	1.57 (0.12)	1.76 (0.28)	1.97 (0.20)
Peak VGRLR (BW/s)	17.65 (3.80)	22.51 (5.44)	27.48 (9.07)	31.56 (7.18)

Conditions: WK00 = unloaded walking, WK15 = 15kg loaded walking, WK25 = 25kg loaded walking, WK35 = 35kg loaded walking. VGRF = vertical ground reaction force. VGRLR = vertical ground reaction loading rate.

No interaction was found between the effects of the subject group and incremented load carriage for the selected gait spatio-temporal parameters and kinetics ($p > 0.05$). In addition, there were no significant differences in gait spatio-temporal parameters and kinetics between the runners and basketball players tested ($p > 0.05$). However, the incremented load carriage had a significant effect on gait spatial-temporal parameters and kinetics ($p < 0.001$).

The incremented load carriage led to a significant increase of stride frequency ($p < 0.001$) and a significant decrease of stride length ($p < 0.001$). Specifically, carrying 35kg load resulted in shorter stride length and faster stride rate than carrying 15kg load and no load conditions ($p < 0.001$). Furthermore, the increasing load carried resulted in significant increases of peak vertical ground reaction force ($p < 0.001$) and peak vertical ground reaction loading rate ($p < 0.001$). As the amount of load carried was increased from 0kg to 35kg, there were linear increases of the peak vertical ground reaction force and peak vertical ground reaction loading rate.

Kinematics and Kinetics of Running Task:

Selected spatio-temporal parameters and ground reaction force variables during running are presented in Table 11. One-way ANOVAs were performed to determine the differences in these variables between the two groups.

Table 11. Means (SDs) spatio-temporal parameters and ground reaction forces during running

Variables\Conditions	Runners	Ball Players
Stride length (m)	2.45 (0.10)	2.46 (0.10)
Stride frequency (strides/min)	82 (3)	82 (3)
Impact peak VGRF (BW)	1.65 (0.05)*	1.81 (0.05)*
Active peak VGRF (BW)	2.43 (0.20)*	2.57 (0.16)*
Peak VGRLR (BW/s)	81.62(22.11)*	95.91(16.12)*

Note: * indicates a significant difference ($P < 0.05$).

VGRF = vertical ground reaction force. VGRLR = vertical ground reaction loading rate.

Both groups exhibited similar stride length ($p = 0.859$) and stride frequency ($p = 0.744$) during running. However, the runner group showed lower ground reaction force variables than those of the basketball group. Specifically, the runner group demonstrated less impact peak VGRF ($p = 0.045$), active peak VGRF ($p = 0.025$), and VGRF loading rate ($p = 0.025$) than those of the basketball group.

Peak Strain and Strain Rate during Loaded Walking Tasks

Strain during walking with load carriages

The following dependent variables were examined: peak tensile strain (peak maximal principle strain), peak compressive strain (peak minimal principle strain), and peak shear strain (peak maximal shear strain) during the stance of walking. For each of the dependent variables, a $2 \times 3 \times 4$ repeated-measures ANOVA test was run with the subject group (two levels: runners and ball players), tibial section (three levels: proximal third, middle third, and distal third), and incremented load carriage (four levels: 0kg, 15kg, 25kg, and 35kg) serving as independent factors. For all the dependent variables examined, significant interactions were found between the subject group and incremented load carriage ($p = 0.0001$) and between the tibial section and incremented load carriage ($p = 0.0001$). Therefore, separate ANOVA tests were run to determine the simple effects from the physical activity history, tibial section, and incremented load carriage on tibial strain variables.

Table 12: Means (SEs) of tibial bone strain during loaded walking tasks

Variables\Conditions	WK00	WK15	WK25	WK35
<u>Runners</u>				
Compressive strain (μs)	658.11 (1.61)	840.41 (1.96)*	924.49 (2.23)*	1011.15 (2.71)*
Tensile strain (μs)	458.33 (1.45)	562.11 (1.81)*	669.82 (2.05)*	733.40 (2.52)*
Shear strain (μs)	1003.29 (2.02)	1229.74 (2.40)*	1444.68 (2.72)*	1586.67 (3.48)*
<u>Basketball Players</u>				
Compressive strain (μs)	634.30 (1.56)	746.87 (1.90)*	842.18 (2.16)*	958.24 (2.63)*
Tensile strain (μs)	440.04 (1.41)	518.86 (1.75)*	597.63 (1.99)*	700.15 (2.47)*
Shear strain (μs)	972.28 (1.96)	1155.32 (2.32)*	1309.96 (2.64)*	1519.01 (3.37)*

Note: data were extracted from the middle third of the tibia shaft. WK00, WK15, WK25 and WK35 represent walking tasks with a load carriage of 0kg, 15kg, 25kg, and 35kg, respectively. * indicates a significant difference ($p = 0.0001$) from the previous level of the loaded walking condition.

One way repeated-measures ANOVAs with the incremented load carriage being the independent factor were run for the runner group and the ball player group, respectively. For both of the groups, significant differences in compressive strain ($p=0.0001$), tensile strain ($p=0.0001$), and shear strain ($p=0.0001$) were found among loaded walking tasks for each of the three tibial sections. Specifically, a dose-response relationship between the incremented load carriage and tibial strain exists. As the load carriage increased in consistent increments, tibial compressive, tensile, and shear strains increase. Table 12 presents the means (SEs) of the compressive, tensile, and shear strains from the middle third of the tibia during walking with incremented load carriages.

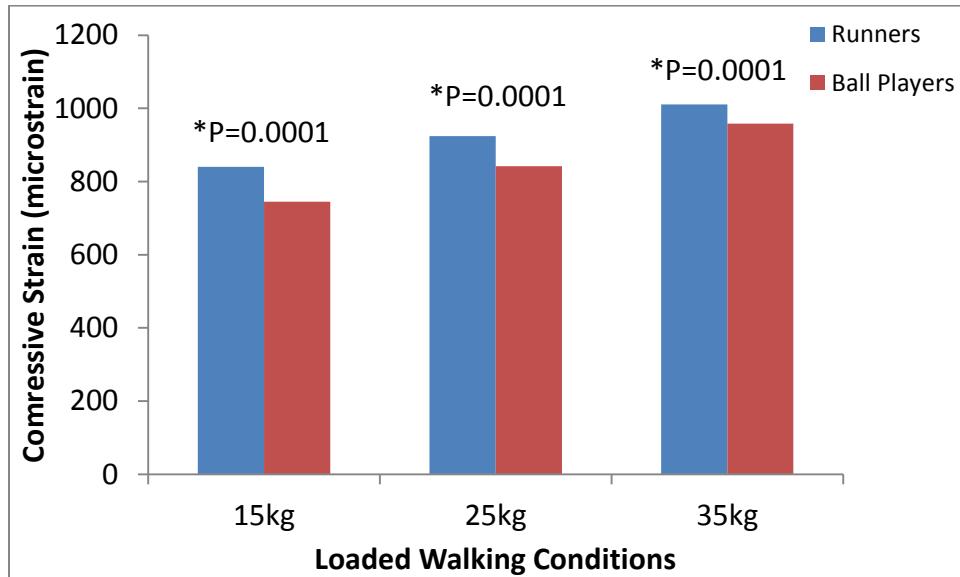


Figure 2: Compressive strain of the middle third of the tibia during loaded walking tasks. The runner group (blue) and the ball player group (red) were compared.

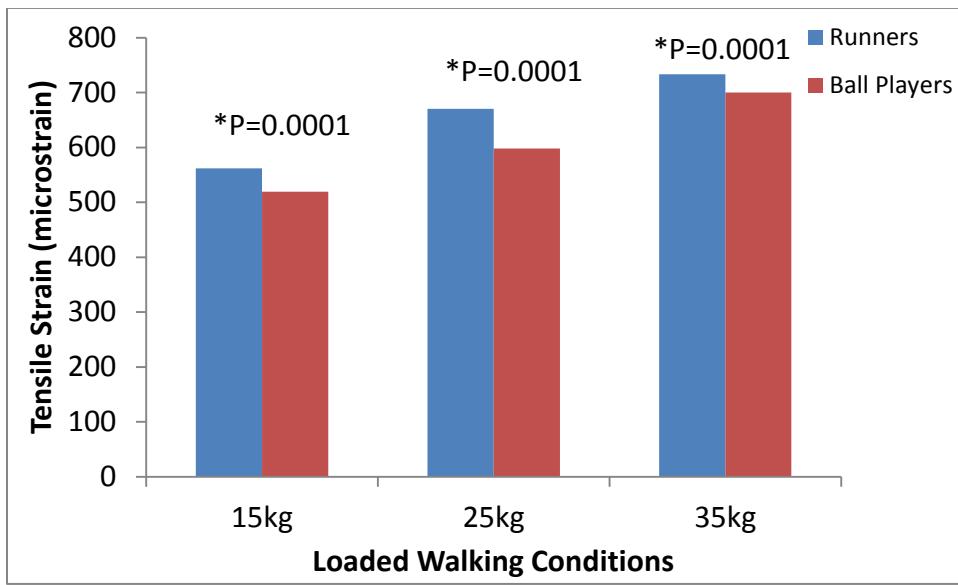


Figure 3: Tensile strains of the middle third of the tibia during loaded walking tasks. The runner group (blue) and the ball player group (red) were compared.

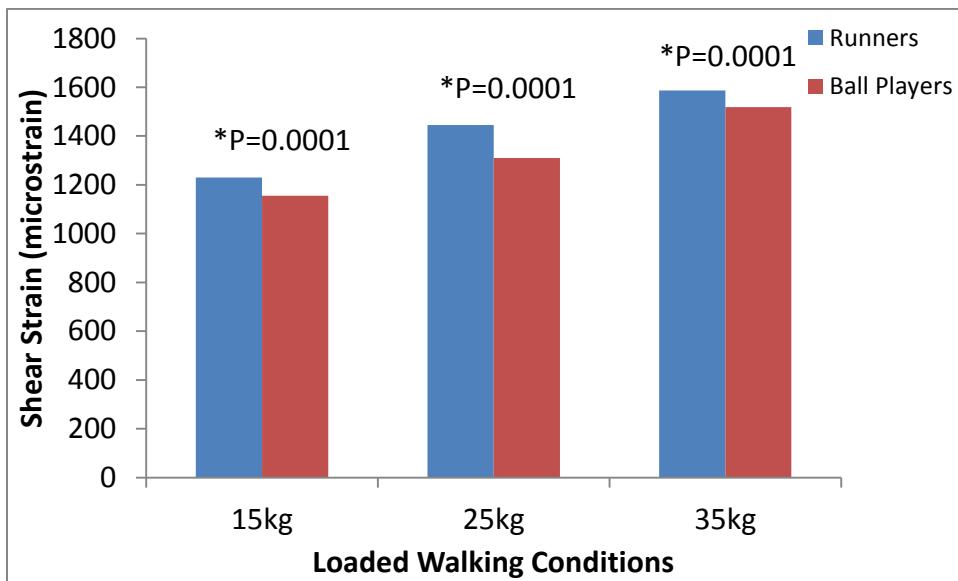


Figure 4: Shear strain of the middle third of the tibia during loaded walking tasks. The runner group (blue) and the ball player group (red) were compared.

One way ANOVA tests with the subject group being the independent factor were run for each of the loaded walking tasks and for each of the tibial sections. Significant differences in compressive strain ($p=0.0001$), tensile strain ($p=0.0001$), and shear strain ($p=0.0001$) were found between the runner group and the ball player group. For all the loaded walking tasks from the 15kg load carriage to 35kg load carriage, the runner group consistently exhibited greater compressive, tensile, and shear strains than the ball player group ($p=0.0001$). The figures 2, 3, and 4 illustrate the group differences in compressive strain, tensile strain, and shear strain during loaded walking tasks, respectively.

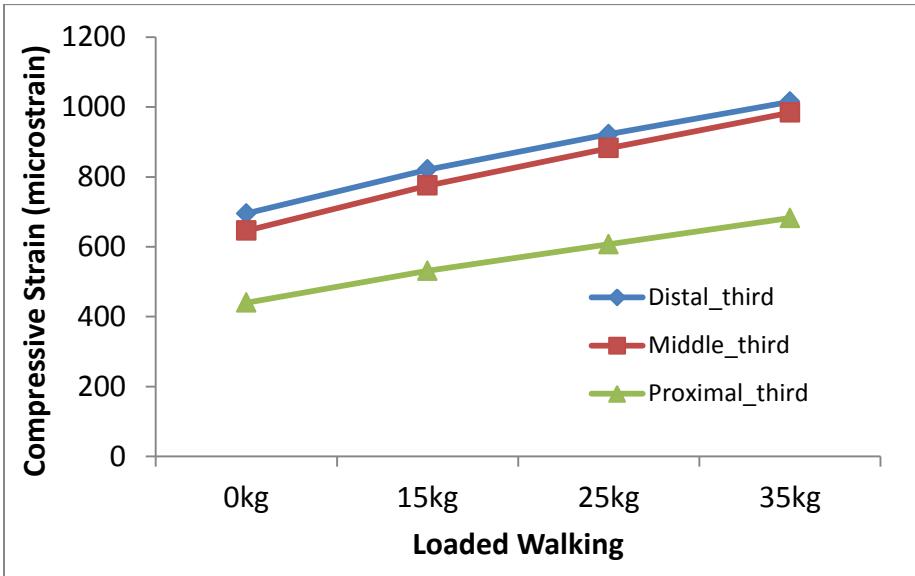


Figure 5: Compressive strain of the proximal, middle, and distal thirds of the tibia during walking with incremented loads.

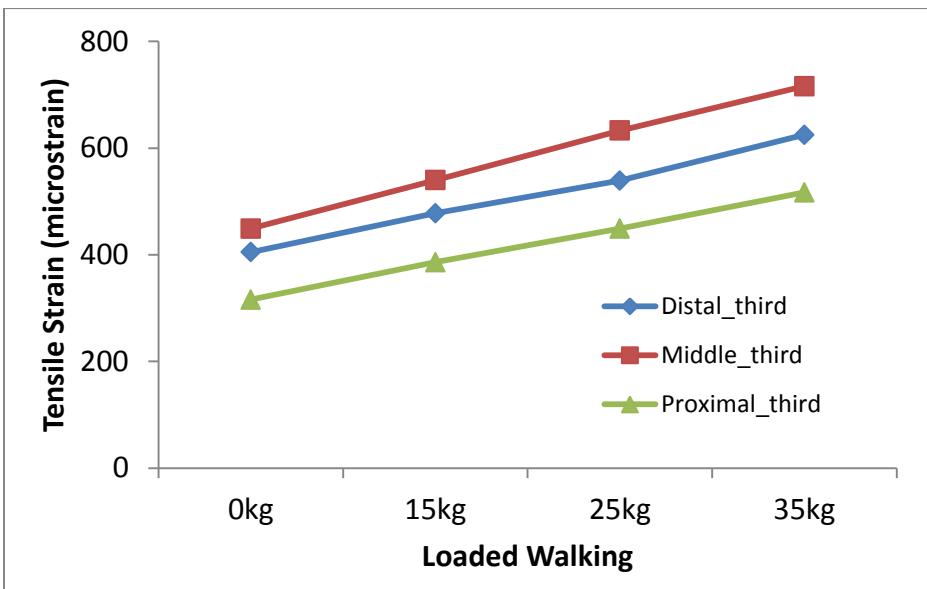


Figure 6: Tensile strain of the proximal, middle, and distal thirds of the tibia during walking with incremented loads.

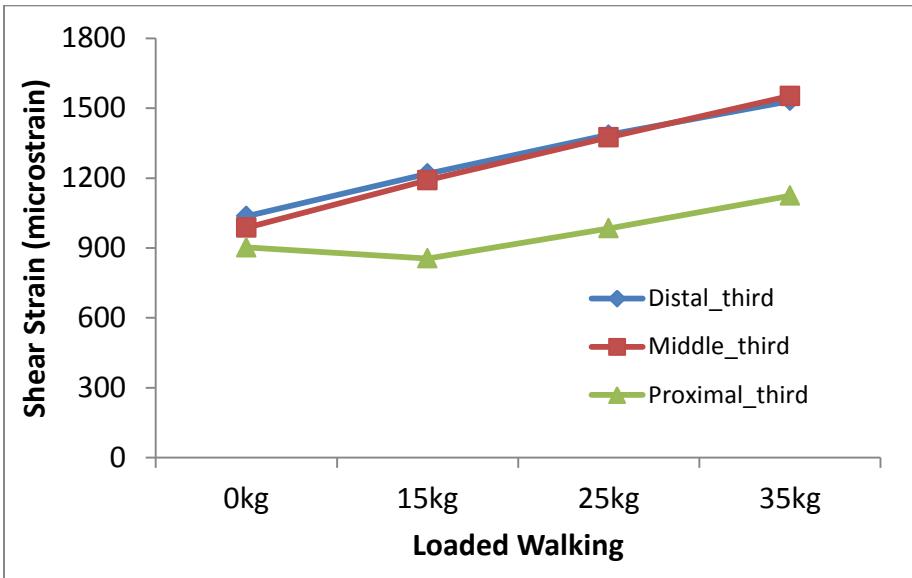


Figure 7: Shear strain of the proximal, middle, and distal thirds of the tibia during walking with incremented loads.

One way ANOVA tests with the tibial section being the independent factor were run for each of the groups and for each of the loaded walking tasks. Significant differences in tibial compressive strain ($p=0.0001$), tensile strain ($p=0.0001$), and shear strain ($p=0.0001$) were found among the three tibia sections. Specifically, for both groups, the distal third of the tibia exhibited greater compressive strain than both of the middle and proximal thirds of the tibia ($p=0.0001$); the middle third of the tibia exhibited greater tensile strain than both of the distal and proximal thirds of the tibia ($p=0.0001$). Furthermore, the ball player group consistently experienced greater shear strain in the distal third of the tibia than in the proximal third of the tibia (during all the loaded walking tasks, $p=0.0001$) and in the middle third of the tibia (during all the loaded walking tasks ($p=0.0001$) except the 35kg condition ($p=0.616$)). The runner group demonstrated greater shear strain in the distal third of the tibia than the other two tibial sections during 0kg and 15kg load carriages ($p=0.0001$). During 25kg load carriage, the runner group experienced similar shear strain in the middle and distal thirds of the tibia ($p=0.602$). During 35kg load carriage, the runner group experienced greater shear strain in the middle third of the tibia than the other two tibial sections ($p=0.0001$). Finally, for both groups, the proximal third of the tibia always experienced the least amount of shear strain within the three tibial sections during loaded walking ($p=0.0001$). Figures 5, 6, 7 illustrate the profiles of the compressive strain, tensile strain, and shear strain of the three tibial sections during various loaded walking tasks, respectively.

Strain during high-impact activities

The following dependent variables were examined: peak tensile strain, peak compressive strain, and peak shear strain during the stance of high-impact activities. For each of the dependent variables, a $2 \times 3 \times 3$ repeated-measure ANOVA test was run with the subject group (two levels: runners and ball players), tibial section (three levels: distal, middle, and proximal thirds of the tibial shaft), and type of activity (three levels: drop-jumping, running, and cutting maneuver) serving as independent factors. For all the dependent variables examined, significant interactions were found between the subject group and type of activity ($p = 0.0001$) and between the tibial section and type of activity ($p=0.0001$). Therefore, separate ANOVA tests were run to determine the simple effects from the physical activity history, tibial section, and high-impact activity on tibial strain variables.

Table 13: Means and SEs of the tibial bone strain during high-impact activities

Variables\Conditions	Running	Jumping	Cutting
<u>Runners</u>			
Compressive strain (μ s)	970.08 (3.56)* \$	1201.59 (5.27)* #	1655.07 (4.37)\$ #
Tensile strain (μ s)	766.23 (2.49)* \$	1058.38 (3.74)* #	1341.02 (3.34)\$ #
Shear strain (μ s)	1627.89 (3.93)* \$	2233.05 (5.64)* #	2906.62 (5.48)\$ #
<u>Basketball Players</u>			
Compressive strain (μ s)	978.96 (3.83)* \$	1316.65 (5.95)* #	1687.21 (4.85)\$ #
Tensile strain (μ s)	736.63 (2.34)* \$	1143.48 (4.06)* #	1351.49 (3.33)\$ #
Shear strain (μ s)	1628.26 (4.34)* \$	2434.92 (6.54)* #	2971.46 (6.04)\$ #

Note: data were extracted from the middle third of the tibia shaft. * indicates a significant difference ($p = 0.0001$) between running and jumping activities; \$ indicates a significant difference ($p = 0.0001$) between running and cutting activities; # indicates a significant difference ($p = 0.0001$) between jumping and cutting activities.

One way repeated-measures ANOVA tests with the high-impact activity being the independent factor were run for each of the group and for each of the tibial sections. Significant differences in compressive strain ($p=0.0001$), tensile strain ($p=0.0001$), and shear strain ($p=0.0001$) were found between the three activities. For both of the subject groups, the cutting maneuver elicited the largest compressive strain ($p=0.0001$), tensile strain ($p=0.0001$), and shear strain ($p=0.0001$) among the three activities across the three tibial sections. The jumping activity produced greater tensile strain ($p=0.0001$) than the running activity for all the tibial sections. Also, the jumping activity produced greater compressive strain ($p=0.0001$) and shear strain ($p=0.0001$) at the middle and proximal thirds of the tibia than the running activity. The running activity only exhibited greater compressive train ($p=0.0001$) and shear strain ($p=0.0001$) than the jumping activity at the distal third of the tibia shaft. Table 13 presents the tibial strains at the middle third of the tibial shaft during the three high-impact activities.

Strain Rate during walking with load carriages:

The following dependent variables were examined: peak tensile strain rate (peak maximal principle strain rate), peak compressive strain rate (peak minimal principle strain rate), and peak shear strain rate (peak maximal shear strain rate) during the stance of walking. For each of the dependent variables, a 2x3x4 repeated-measure ANOVA test was run with the subject group (two levels: runners and ball players), tibial section (three levels: proximal third, middle third, and distal third), and incremented load carriage (four levels: 0kg, 15kg, 25kg, and 35kg) serving as independent factors. For all the dependent variables examined, significant interactions were found between the subject group and incremented load carriage ($p = 0.0001$) and between the tibial section and incremented load carriage ($p = 0.0001$). Therefore, separate ANOVA tests were run to determine the simple effects from the physical activity history, tibial section, and incremented load carriage on tibial strain rate variables.

Table 14: Means (SEs) of tibial bone strain rate during loaded walking tasks

Variables\Conditions	WK00	WK15	WK25	WK35
<u>Runners</u>				
Compressive strain rate ($\mu\text{s}/\text{s}$)	7955.99 (24.08)	9477.93 (27.04)*	11562.32 (33.26)*	11583.09 (37.50)
Tensile strain rate ($\mu\text{s}/\text{s}$)	6647.40 (20.07)	7901.71 (23.27)*	9551.94 (28.28)*	9617.21 (33.14)\$
Shear strain rate ($\mu\text{s}/\text{s}$)	6668.60 (13.30)	7828.50 (15.27)*	9594.68 (20.05)*	9662.64 (24.26)\$
<u>Basketball Players</u>				
Compressive strain rate ($\mu\text{s}/\text{s}$)	7882.92 (23.34)	9034.16 (26.22)*	10293.25 (32.23)*	11661.34 (36.36)*
Tensile strain rate ($\mu\text{s}/\text{s}$)	6557.76 (19.46)	7432.32 (22.57)*	8632.39 (27.42)*	10012.88 (32.13)*
Shear strain rate ($\mu\text{s}/\text{s}$)	6540.35 (12.90)	7488.80 (14.80)*	8609.44 (19.44)*	9976.34 (23.53)*

Note: data were extracted from the middle third of the tibia shaft. WK00, WK15, WK25 and WK35 represent walking tasks with a load carriage of 0kg, 15kg, 25kg, and 35kg, respectively. Significant differences between the current load carriage condition and previous load carriage condition were indicated (* indicates $p=0.0001$; \$ indicates $p<0.05$).

One way repeated-measures ANOVAs with the incremented load carriage being the independent factor were run for the runner group and the ball player group, respectively. For both of the groups, significant differences in compressive strain rate ($p=0.0001$), tensile strain rate ($p=0.0001$), and shear strain rate ($p=0.0001$) were found among loaded walking tasks for each of the three tibial sections. Table 14 presents the means (SEs) of the compressive, tensile, and shear strains rate from the middle third of the tibia during walking with incremented load carriages.

Compressive strain rate was evaluated for both of the subject groups and for each of the tibial sections. For the runner group, the distal third of the tibia exhibited pronounced increases of the compressive strain rate as the load carriage increased from 0kg to 15kg ($p=0.0001$), from 15kg to 25kg ($p=0.0001$), and from 25kg to 35kg ($p=0.06$); the middle third of the tibia exhibited significant increases of the compressive strain rate as the load carriage increased from 0kg to 15kg ($p=0.0001$) and from 15kg to 25kg ($p=0.0001$); meanwhile, the proximal third of the tibia exhibited significant increases in the compressive strain rate as the load carriage increased from 0kg to 15kg ($p=0.0001$), from 15kg to 25kg ($p=0.0001$) and from 25kg to 35kg ($p=0.0001$). For the ball player group, all the three tibial sections exhibited significant increases of compressive strain rate as the load carriage increased from 0kg to 15kg ($p=0.0001$), from 15kg to 25kg ($p=0.0001$), and from 25kg to 35kg ($p=0.0001$).

Tensile strain rate was evaluated for both of the subject groups and for each of the tibial sections. For the runner group, the distal third of the tibia exhibited significant increases of the tensile strain rate as the load carriage increased from 0kg to 15kg ($p=0.0001$) and from 15kg to 25kg ($p=0.0001$); the middle third of the tibia exhibited significant increases of the tensile strain rate as the load carriage increased from 0kg to 15kg ($p=0.0001$), from 15kg to 25kg ($p=0.0001$), and from 25kg to 35kg ($p=0.033$); the proximal third of the tibia also exhibited significant increases of the tensile strain rate when the load carriage increased from 0kg to 15kg ($p=0.0001$), from 15kg to 25kg ($p=0.0001$), and from 25kg to 35kg ($p=0.0001$). For the ball player group, all the three tibial sections exhibited significant increases of the tensile strain rate as the load carriage increased from 0kg to 15kg ($p=0.0001$), from 15kg to 25kg ($p=0.0001$), and from 25kg to 35kg ($p=0.0001$), and from 25kg to 35kg ($p=0.0001$).

Shear strain rate was evaluated for both of the subject groups and for each of the tibial sections. For the runner group, the distal third of the tibia exhibited significant increases of the shear strain rate as the load carriage increased from 0kg to 15kg ($p=0.0001$) and from 15kg to 25kg ($p=0.0001$); the middle third of the tibia demonstrated significant increases of the shear strain rate as the load carriage increased from 0kg to 15kg ($p=0.0001$), from 15kg to 25kg ($p=0.0001$), and from 25kg to 35kg ($p=0.01$); meanwhile, the proximal third of the tibia demonstrated significant increases of the shear strain rate as the load carriage increased from 0kg to 15kg ($p=0.0001$), from 15kg to 25kg ($p=0.0001$), and from 25kg to 35kg ($p=0.0001$). For the ball player group, all the three tibial sections exhibited significant increases of the shear strain rate as the load carriage increased from 0kg to 15kg ($p=0.0001$), from 15kg to 25kg ($p=0.0001$), and from 25kg to 35kg ($p=0.0001$).

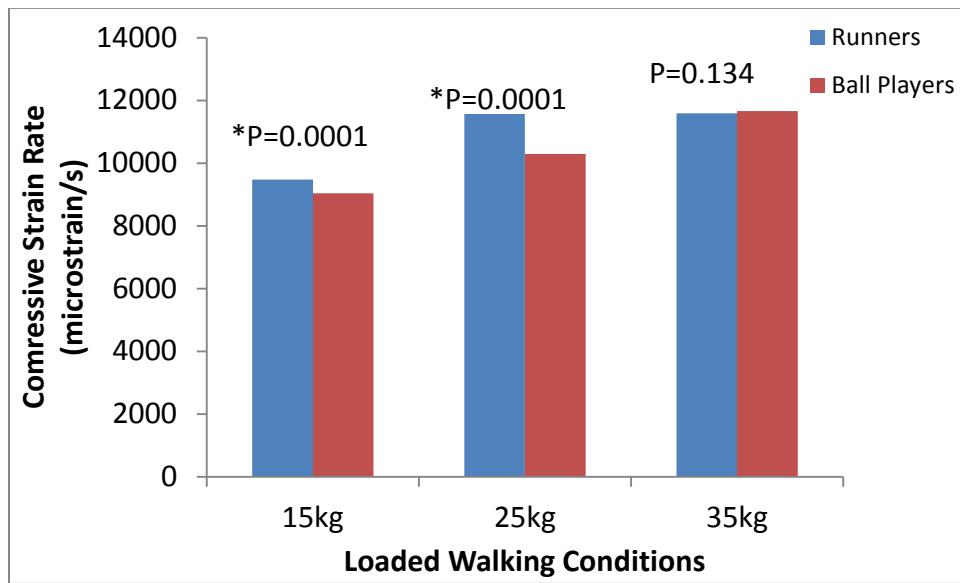


Figure 8: Compressive strain rate of the middle third of the tibia during loaded walking tasks. The runner group (blue) and the ball player group (red) were compared.

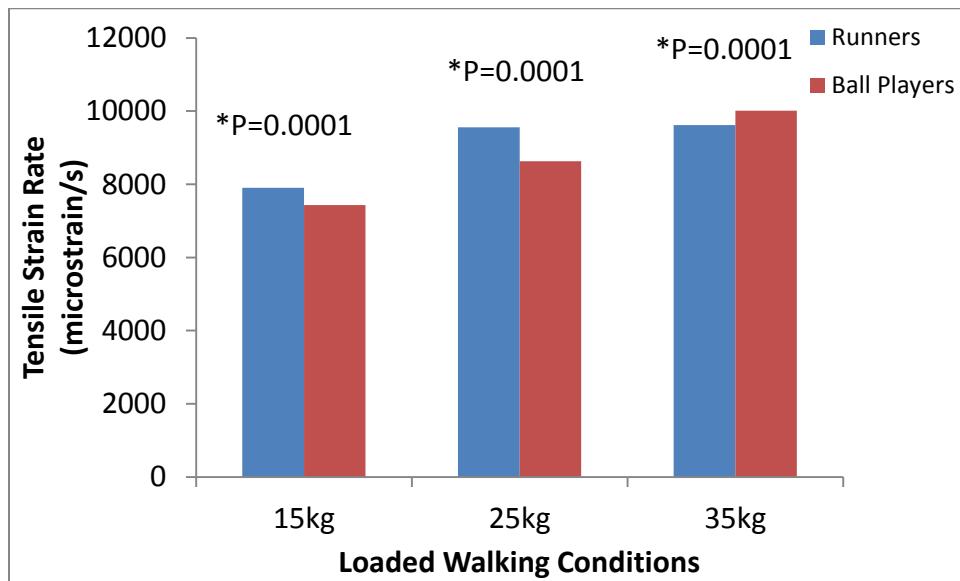


Figure 9: Tensile strain rate of the middle third of the tibia during loaded walking tasks. The runner group (blue) and the ball player group (red) were compared.

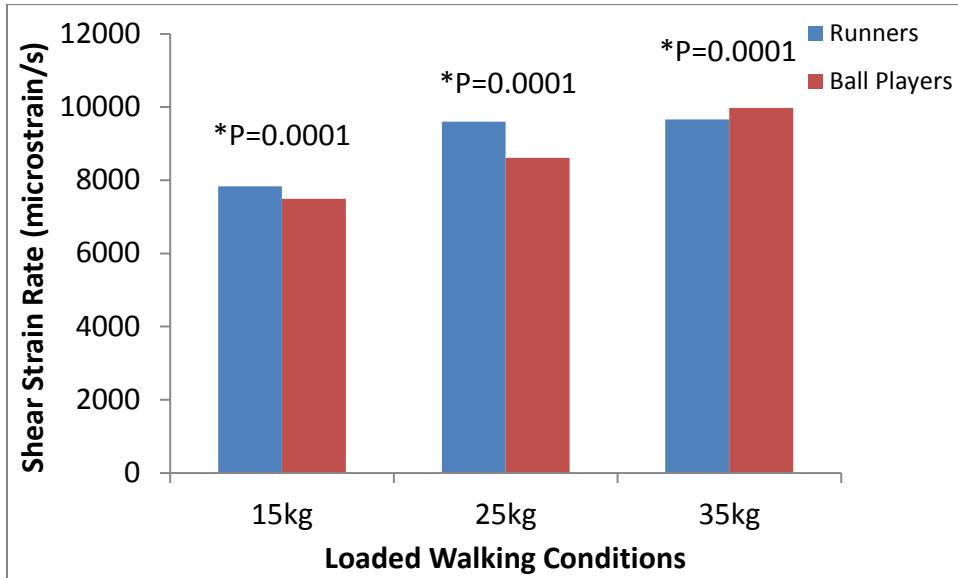


Figure 10: Shear strain rate of the middle third of the tibia during loaded walking tasks. The runner group (blue) and the ball player group (red) were compared.

One way ANOVA tests with the subject group being the independent factor were run for each of the loaded walking tasks and for each of the tibial sections. Significant differences in compressive strain rate ($p=0.0001$), tensile strain rate ($p=0.0001$), and shear strain rate ($p=0.0001$) were found between the runner group and the ball player group. Figures 8, 9, and 10 illustrate the group differences in compressive strain rate, tensile strain rate, and shear strain rate during loaded walking tasks, respectively.

The compressive strain rate was evaluated for each of the loaded walking tasks and for each of the tibial sections. During the 15kg and 25kg load carriages, the runner group demonstrated greater strain rate than the ball player group ($p=0.0001$) across all tibial sections. During the 35kg load carriage, compared to the ball player group, the runner group exhibited less strain rate ($p=0.005$) at the distal third of the tibia, more strain rate ($p=0.0001$) at the proximal third of the tibia, and no difference ($p=0.134$) in strain rate at the middle third of the tibia.

The tensile strain rate was evaluated for each of the loaded walking tasks and for each of the tibial sections. During the 15kg and 25kg load carriages, the runner group demonstrated greater strain rate than the ball player group ($p=0.0001$) across all tibial sections. During the 35kg load carriage, compared to the ball player group, the runner group exhibited less tensile strain rate ($p=0.0001$) at the distal third and middle third of the tibia and more strain rate ($p=0.0001$) at the proximal third of the tibia.

The shear strain rate was evaluated for each of the loaded walking tasks and for each of the tibial sections. During the 15kg and 25kg load carriages, the runner group demonstrated greater strain rate than the ball player group ($p=0.0001$) across all tibial sections. During the 35kg load carriage, compared to the ball player group, the runner group exhibited less shear strain rate ($p=0.0001$) at the distal third and middle third of the tibia and more strain rate ($p=0.0001$) at the proximal third of the tibia.

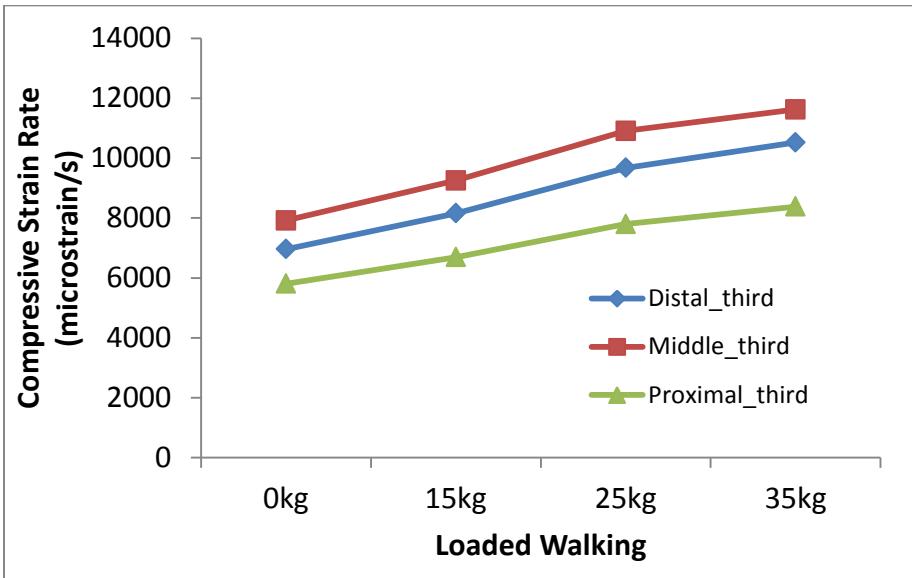


Figure 11: Compressive strain rate of the proximal, middle, and distal thirds of the tibia during walking with incremented loads.

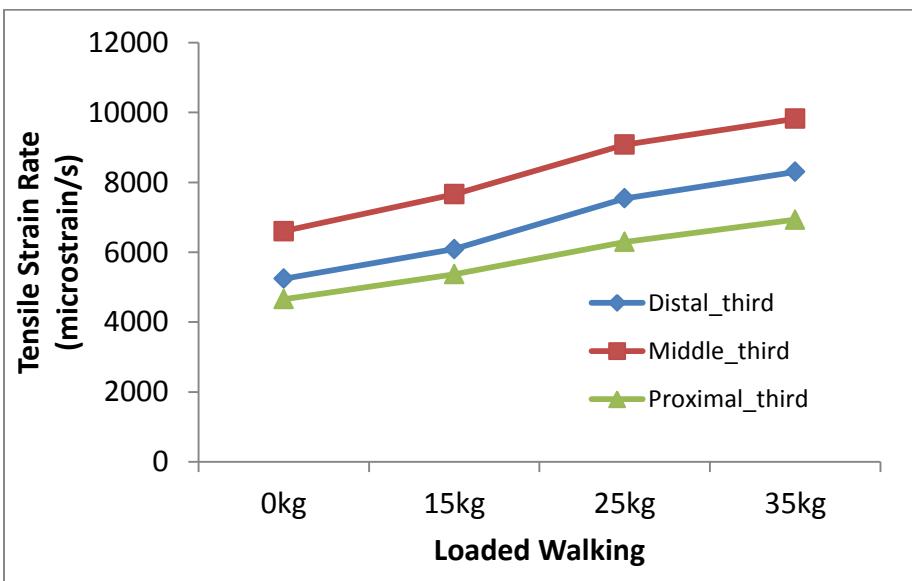


Figure 12: Tensile strain rate of the proximal, middle, and distal thirds of the tibia during walking with incremented loads.

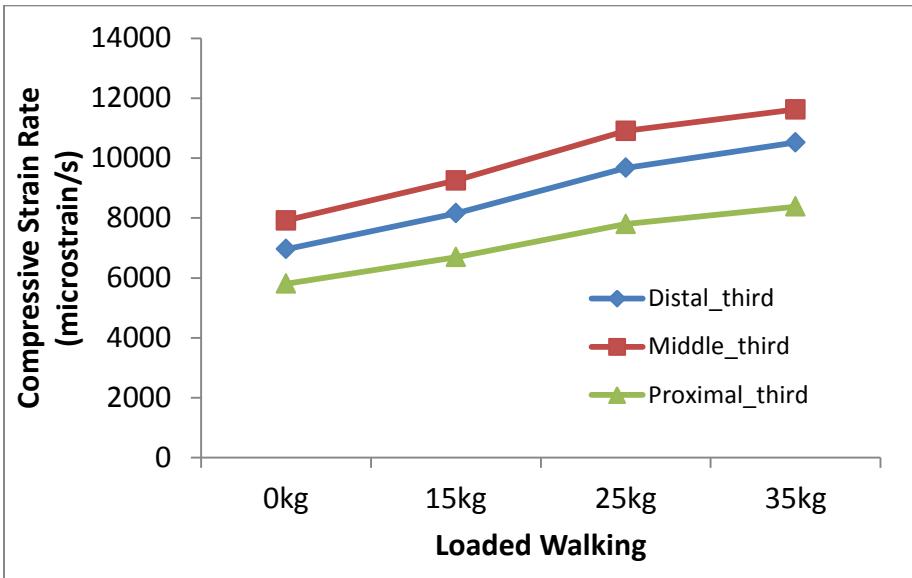


Figure 13: Compressive strain rate of the proximal, middle, and distal thirds of the tibia during walking with incremented loads.

One way ANOVA tests with the tibial section being the independent factor were run for each of the groups and for each of the loaded walking tasks. Significant differences in tibial compressive strain rate ($p=0.0001$), tensile strain rate ($p=0.0001$), and shear strain rate ($p=0.0001$) were found among the three tibia sections. Specifically, for both groups, the middle third of the tibia exhibited the greatest compressive strain rate ($p=0.0001$), tensile strain rate ($p=0.0001$), and shear strain rate ($p=0.0001$) among the three tibial sections across all load carriages. In addition, the proximal third of the tibia exhibited the least amount of strain rates in compression ($p=0.0001$), tension ($p=0.0001$), and shear ($p=0.0001$) among the three tibial sections across all load carriages. Figures 11, 12, 13 illustrate the profiles of the compressive strain rate, tensile strain rate, and shear strain rate of the three tibial sections during various loaded walking tasks, respectively.

Strain rate during high-impact activities

The following dependent variables were examined: peak tensile strain rate, peak compressive strain rate, and peak shear strain rate during the stance of high-impact activities. For each of the dependent variables, a $2 \times 3 \times 3$ repeated-measure ANOVA test was run with the subject group (two levels: runners and ball players), tibial section (three levels: distal, middle, and proximal thirds of the tibial shaft), and type of activity (three levels: drop-jumping, running, and cutting maneuver) serving as independent factors. For all the dependent variables examined, significant interactions were found between the subject group and type of activity ($p = 0.0001$) and between the tibial section and activity ($p=0.0001$). Therefore, separate ANOVA tests were run to determine the simple effects from the physical activity history, tibial section, and high-impact activity on tibial strain rate variables.

Table 15: Means and SEs of the tibial bone strain rate during high-impact activities

Variables\Conditions	Running	Jumping	Cutting
<u>Runners</u>			
Compressive strain rate ($\mu\text{s}/\text{s}$)	17593.96 (69.08)* \$	18021.50 (92.58)* #	28779.64 (87.93)\$ #
Tensile strain rate ($\mu\text{s}/\text{s}$)	14954.92 (59.08)* \$	16705.46 (73.54)* #	23193.99 (68.85)\$ #
Shear strain rate ($\mu\text{s}/\text{s}$)	15314.35 (42.10)* \$	17029.96 (56.36)* #	24016.01 (49.17)\$ #
<u>Basketball Players</u>			
Compressive strain rate ($\mu\text{s}/\text{s}$)	17246.25 (69.08)* \$	19367.54 (102.68)* #	27750.49 (90.16)\$ #
Tensile strain rate ($\mu\text{s}/\text{s}$)	14023.33 (58.65)* \$	18040.32 (80.97)* #	22490.60 (65.28)\$ #
Shear strain rate ($\mu\text{s}/\text{s}$)	14709.81 (44.30)* \$	18435.83 (64.55)* #	23682.54 (52.53)\$ #

Note: data were extracted from the middle third of the tibia shaft. * indicates a significant difference ($p = 0.0001$) between running and jumping activities; \$ indicates a significant difference ($p = 0.0001$) between running and cutting activities; # indicates a significant difference ($p = 0.0001$) between jumping and cutting activities.

One way repeated-measures ANOVA tests with the high-impact activity being the independent factor were run for each of the group and for each of the tibial sections. Significant differences in compressive strain rate ($p=0.0001$), tensile strain rate ($p=0.0001$), and shear strain rate ($p=0.0001$) were found between the three activities. For both of the subject groups, the cutting maneuver elicited the largest compressive strain rate ($p=0.0001$), tensile strain rate ($p=0.0001$), and shear strain rate ($p=0.0001$) among the three activities across the three tibial sections; the jumping activity produced greater compressive strain rate ($p=0.0001$), tensile strain rate ($p=0.0001$), and shear strain rate ($p=0.0001$) than the running activity at the middle and proximal thirds of the tibia; the running activity only led to greater compressive strain rate ($p=0.0001$) and shear strain rate ($p=0.0001$) at the distal third of the tibia than the jumping activity. Finally, running activity performed by the runner group exhibited greater tensile strain rate ($p=0.0001$) at the distal third of the tibia than the jumping activity, while the running activity performed by the ball player group exhibited less tensile strain rate ($p=0.0001$) at the distal third of the tibia than the jumping activity. Table 15 presents the tibial strain rates at the middle third of the tibial shaft during the three high-impact activities.

SUMMARY and DISCUSSION:

Load carriage leads to significant alterations of gait spatio-temporal parameters and kinetics. The increases of the vertical ground reaction force and loading rate are proportional to the increments of load carried. Although runners and basketball players recruited in this study had distinct physical activity histories, they exhibited no differences in ground reaction forces and loading rates during loaded walking tasks. The results imply that when participating in unaccustomed activities (e.g. loaded walking), both the runner and basketball groups could experience similar mechanical loadings.

When performing running movement, habitual runners recruited in this study demonstrated an ability to lower the ground reaction forces. Running with reduced ground impact forces could result in decreased risks of impact related overuse injuries. As the runners have adapted to the running environment by lowering impact forces, runners' tibia bones may also have adapted to the same loading environment.

On one hand, adaptation to a loading environment with reduced impact forces could mean the tibia bone may have remodeled itself for the reduced imposed mechanical loadings via decreasing bone density at some parts of the bone for metabolic efficiency purposes. On the other hand, when the loading environment is changed to an unaccustomed situation (e.g. loaded walking), the tibia bone may need to accelerate the remodeling process to become capable of resisting the novel loadings.

We had hypothesized that the incremented load carriages would lead to increases of tibia bone strain. This hypothesis was supported. It was found that the tibial compressive, tensile, and shear strains increased significantly as the load carriage increased from 0kg to 15kg, from 15kg to 25kg and from 25kg to 35kg. The current finding of this significant dose-response relationship between the load carriage and tibial strain indicates that the tibia bone deformation during loaded walking is positively related to the amount of load carried. Thus, it is possible that increasing the load carriage during military training could lead to increased risk of tibia bone damage due to the increase of bone deformation.

We had hypothesized that the past physical-training history could influence the tibial bone strain during loaded walking. This hypothesis was supported. We found the runner group demonstrated larger tibial compressive, tensile, and shear strain than the ball player group during walking with incremented load carriages. Participating in running for minimum of two years, the runner group may have remodeled their tibial bones to accommodate the uni-axial cyclic loading encountered during running. On the contrary, minimum of two years of experiencing the irregular multi-axial loading during basketball may shape the tibias of the ball player group to better resist unaccustomed loadings. In this study, the tibias of the ball player group demonstrated less bone deformation during loaded walking than those of the runner group. Thus, it is evident that the tibias of the ball player group are more resilient to the unaccustomed loading from load carriages than those of the runner group.

The strains of the proximal, middle, and distal thirds of the tibia were examined during load carriages. It was found that the middle third of the tibia always experienced the largest tensile strain and the distal third of the tibia always experienced the largest compressive strain among the three tibial thirds. The proximal third of the tibia was found to experience the least amount of the strains among the three tibial thirds during load carriages. Previous studies examined *in vivo* tibial bone strain at the anterior-medial aspect of the middle bone shaft and considered this site was a common location of tibial stress fracture [13, 14, 17]. Findings from this study provide evidence to the assumption taken by those studies [13, 14, 17]. As bone is generally weak to tensile stress, sustaining pronounced tensile strain at the middle third of the tibia during loaded walking could result in increased risk of stress fractures.

We had hypothesized that high impact, irregular sporting maneuvers such as cutting and drop-jumping, which were consistent with those performed in basketball, produce different tibial strain patterns than those produced during running. This hypothesis was supported. The cutting maneuver elicited largest amount of compressive, tensile, and shear strains among the three high-impact activities tested. The drop-jumping activity produced greater strains at the middle third of the tibia than the running activity. The running activity only resulted in larger compressive strain than drop-jumping at the distal third of the tibia. Thus, it is evident that participating in high-impact irregular maneuvers consisting cutting and drop-jumping could lead to strengthening tibial bone in general; in particular, the bone quality of the middle third of the tibia can be improved to become more resilient to tensile loads. As the middle third of the tibia is a vulnerable site sustaining stress fracture during military training [13, 14, 17], a pre-conditioning program consisting high-impact irregular activities such as cutting and drop-jumping may be beneficial to military recruits who are entering basic training and facing high risk of tibial stress fractures.

We had hypothesized that incremented load carriages would result in increases of tibial bone strain rates. This hypothesis was supported in general. We found the ball player group exhibited pronounced increases of compressive, tensile, and shear strain rates at all the three tibial sections as the load carriage increased from 0kg to 15kg, from 15kg to 25kg, and from 25kg to 35kg, respectively. The runner group also experienced similar strain rate patterns when the load carriage increased from 0kg to 15kg and from 15kg to 25kg. In addition, the runner group experienced the following strain rate patterns: increased compressive strain rate at the distal third of the tibia with the load carriage increasing from 25kg to 35kg; increased tensile strain rate and shear strain rate at the middle third of the tibia with the load carriage increasing from 25kg to 35kg. Interestingly, the compressive strain rate at the middle third of the tibia, the tensile strain rate and shear strain rate at the distal third of the tibia stay unchanged with the load carriage increasing from 25kg to 35kg. As load carriage is accompanied with alterations of gait patterns [4, 18], it is possible that the adjustments of gait signified by reducing stride length and increasing lower-extremity flexion angles could have an effect on strain rate. Reducing the strain rate at the 25kg level of load carriage may be a response from the runner group to prevent bone damage due to cyclic loading during load carriage. On the contrary, the ball player group did not exhibit a threshold effect on bone strain rate before the 35kg load. This reflects that the tibias of the ball player group may be strong enough to sustain higher level loads than the 35kg load.

We had hypothesized that the past physical-training history could influence the tibial bone strain rate during loaded walking. This hypothesis was supported. The runner group consistently demonstrated larger tibial strain rates than those of the ball player group when carrying 15kg and 25kg loads. Interestingly, when the load carriage increased to a 35kg load, the runner group started showing lower strain rates than the ball player group. It is possible that a strain rate threshold exists during loaded walking. This strain rate threshold may be dependent on the tibial bone quality. If the tibial bone is not accustomed to a loading environment such as incremented load carriages in this study, protective strategies signified by alternating gait kinematics may result in a reduction of bone strain rates. Therefore, as the runner group experienced unchanged strain rates after the 25kg load carriage and lower strain rates at 35kg load carriage than the ball player group, it is likely that the tibias of the runner group may be less resilient to load carriages than the tibias of the ball player group.

When examining the strain rates at the proximal, middle, and distal thirds of the tibia during load carriages, we found the middle third of the tibia sustained the highest strain rate within the three tibial sections. The middle third of the tibia has long been considered a vulnerable site sustaining stress fracture during military training [13, 14, 17]. Findings from this study provide evidence to support this consensus. During military basic training, it is possible that load carriage could lead to a pronounced increase of strain rates at the middle third of the tibia. Bone remodeling may be accelerated, which could temporarily weaken the bone structure. If the load carriage training continues, the risk of developing stress fracture at the middle third of the tibia could be increased.

We had hypothesized that high impact, irregular sporting maneuvers such as cutting and drop-jumping, which were consistent with those performed in basketball, produce different tibial strain rate patterns than those produced during running. This hypothesis was supported. We found that the cutting maneuver generated the highest strain rates than the drop-jumping and running activities and the drop-jumping resulted in greater strain rates at the middle third of the tibia than running. Thus, it is evident that participating in sports comprising high impact multi-directional loadings such as cutting and drop-jumping could be beneficial to strengthen tibial bones, which could become resilient to load carriages.

CONCLUSION

Load carriages result in increases in tibial strains and strain rates. A dose-response relationship exists between incremented load carriages and bone strains. A threshold in strain rate may exist during walking with incremented load carriages. This threshold may serve as a protective mechanism to limit the increase of strain rates. Past training history has an influence on tibial strains and strain rates. For any given load carriage, basketball players experienced smaller increases of bone strains than runners. During load carriages, the middle third of the tibia experienced the highest strain rates and tensile strain among the three tibial sections. Cutting and drop-jumping maneuvers demonstrated greater strain and strain rates at the middle third of the tibia than running. For military recruits entering the basic training, a pre-conditioning program comprised with high impact, irregular sporting maneuvers such as basketball may be beneficial for strengthening tibial bones to resist unaccustomed loadings from load carriage training.

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Appendix A

Abstract presented at the North West Biomechanics Symposium (NWBS) in Seattle, WA. May 21 – May 22, 2010.

Title: An Integrated Modeling Method for Tibia Strain Analysis

AN INTEGRATED MODELING METHOD FOR TIBIA STRAIN ANALYSIS

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INTRODUCTION

Stress fracture is a common overuse injury experienced during activities involving repetitive loading (running and loaded walking). A common location for these fractures is the tibia [1]. The cause of tibial stress fracture remains elusive despite considerable research [2]. A reason for ambiguity may be the use of surrogate variables instead of the mechanical causes of bone microdamage (deformation). Gathering bone strain data *in vivo* has inherent limitations, so it is useful to explore musculo-skeletal modeling techniques that can assess changes in bone strain response across individuals. The goal of this project was to develop methods to generate strain data for large cohorts of subjects.

METHODS

Seventeen college-aged male subjects participated. Walking kinematic and kinetic data were collected for unloaded walking at 1.67 m/s on an AMTI force instrumented treadmill (AMTI, Watertown, MA). Bone geometries were obtained using computed tomography (CT). CT images were segmented and 3D geometry files generated in Materialise MIMICS 13.0 (Materialise, Leuven, Belgium).

The 3D surface geometries were used in MD MARC 2008 (MSC.Software, Santa Anna, CA) to build hexmesh finite element (FE) models with generic linear isotropic material properties of elastic modulus 17GPa, density 1.9g/cm³, and Poisson's ratio of 0.3. Tibia models were then imported into a scaled LifeMOD model. Once positioned, spatial coordinates of muscle model markers representing origins, insertions, and joint positions relative to the tibia and incorporated into the FE tibia. Boundary conditions were assigned as rotational and translational degrees of freedom of nodes representing ankle and knee joint centers. Flexible bodies (modal neutral files) were generated for LifeMOD. For a full description of LifeMOD modeling methods and use of flexible bodies, see Al Nazer [3]. Key differences between this study and that of Al Nazer et al include a more muscle actuators on the leg that includes the flexible tibia, subject specific segment scaling based on joint center calculations, ground reaction forces, and subject specific tibial geometries generated from CT scans.

A lower body model was built using subject sex, mass, and height as scaling inputs in the GeBOD [4] database. Segments, joint locations, and orientations were scaled

using joint center data from Visual3D. Kinematic data were used to perform an inverse kinematic analysis. Results of this analysis were used to "train" the muscle (right leg) and joint (left leg) PID controllers used to actuate the model in a forward dynamics (FD) analysis. Flexible tibias were then imported into LifeMOD. A FD analysis was performed with the addition of ground reaction forces applied to the feet and motion capture kinematics disabled. Maximum principle, minimum principle, and maximum shear strain values were then calculated using the Durability plug-in for MD ADAMS/View for the nodes of the geometric middle third of the tibial shaft [3,5] (MSC. Software, Santa Ana, CA).

RESULTS AND DISCUSSION

Results are in Table 1. Maximum principle and minimum principle strains and strain rates appear to be consistent with previously reported values, though maximum shear strain and strain rate values are lower. Further analysis is needed to determine if these differences are due to modeling methods or to differences in experimental protocols, specifically treadmill versus over-ground walking.

Methods used in this study provide a promising means of investigating tibial strain and strain rate across individuals and conditions. Since these methods do not require bone staples, more diverse populations can be studied in the future.

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ACKNOWLEDGEMENTS

Funding source: Department of the Army #W81XWH-08-1-0587

Ed Rezer of Device Analytics for FE analysis support

Table 1: Averaged Strains and Strain Rate Comparisons

	Strain (Microstrain)			Strain Rate (Microstrain/s)		
	Max Prin	Min Prin	Max Shear	Max Prin Rate	Min Prin Rate	Max Shear Rate
Al Nazer et al ³	305	-645	948	4000	-7000	10000
Present Simulation	400	-603	463	3168	-4578	3240

Appendix B

Abstract presented at the 56th annual meeting of American College of Sports Medicine (ACSM) in Baltimore, MD. June 2- June 5, 2010.

The abstract was published in *Medicine & Science in Sports & Exercise*. V. 42, No. 5 Supplement, S192

Title: Muscular Fatigue Increases Ground Reaction Loading Rate during Walking

Medicine & Science in Sports & Exercise:
May 2010 - Volume 42 - Issue 5 - pp 271-272
doi: 10.1249/01.MSS.0000384356.85597.85

A-24 Free Communication/Poster - Gait Analysis: JUNE 2, 2010 7:30 AM - 12:30 PM: ROOM: Hall C

Muscular Fatigue Increases Ground Reaction Loading Rate During Walking: 1433: Board #89 June 2 11:00 AM - 12:30 PM

WANG, HE¹; FRAME, JEFF¹; OZIMEK, ELICIA¹; REEDSTROM, CARA¹; LEIB, DANIEL²; DUGAN, ERIC²

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(*No disclosure reported*)

Military personnel are commonly afflicted by overuse injuries such as tibia stress fracture during basic physical training. Muscular fatigue is thought to reduce the leg muscles' ability to attenuate dynamic load on human musculoskeletal system during locomotion (Voloshin, et al. 1998). However, it is not clear if or how muscular fatigue influences mechanical loading of the lower extremities and increases the risk of developing tibia stress fracture.

PURPOSE: Compare peak ground impact forces and loading rates during a fatigued and unfatigued walking task.

METHODS: Fourteen healthy male subjects (age: 21 ± 2 yr.; body mass: 81.5 ± 11.3 kg; body height: 182 ± 4 cm) participated in the study. Subjects participated in a fatiguing protocol which involved a series of metered step-ups and heel raises while wearing a 16 kg rucksack. The presence of fatigue was determined by a decline in performance of a vertical jump test (< 80% of vertical jump max) Prior to and immediately after the fatiguing protocol, subjects performed level walking at 1.67 m/s on a force instrumented treadmill (AMTI). The following variables were analyzed: Peak vertical and antero-posterior ground reaction forces and peak instantaneous vertical and braking loading rates during first half of the stance. A one way repeated measures MANOVA was used to determine differences in these variables between unfatigued and fatigued conditions. $\alpha = 0.05$.

RESULTS: Compared to unfatigued condition, during the fatigued condition the subjects exhibited greater vertical ground reaction forces (1.36 ± 0.11 vs. 1.30 ± 0.07 BW, respectively) ($p = 0.000$), greater instantaneous vertical loading rates (24.46 ± 9.92 vs. 18.28 ± 4.51 BW/s, respectively.) ($p = 0.000$), and greater instantaneous braking loading rates (-10.48 ± 3.02 vs. -8.61 ± 1.99 BW/s, respectively.) ($p = 0.001$).

CONCLUSION: Increased vertical loading rate has been linked to higher risk of tibia stress fracture for distance runners (Milner et al. 2006). In this study, muscular fatigue seems to lead to increases of vertical ground impact and instantaneous loading rates during walking at the given speed. It is possible that military personnel under the influence of muscular fatigue may face increased risk of tibia stress fracture when participating in intensive physical training programs. *DoD#W81XWH-08-1-0587*

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Appendix C

Abstract presented at the 34th Annual Meeting of American Society of Biomechanics (ASB) in
Provincial, RI, August 18- August 20, 2010

Title: Load Carriage Increases Mechanical Loading Rates during Walking

LOAD CARRIAGE INCREASES MECHANICAL LOADING RATES DURING WALKING

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INTRODUCTION

Stress fracture is an overuse bone injury. It is a result of excessively repetitive loads acting on the bone over time which leads to fatigue induced bone microdamage [1,2]. Mechanical loading and loading rate are two major factors related to stress fracture development.

Military personnel are commonly afflicted by stress fractures during basic training. The injury rate of stress fractures during basic training is approximately 6% in the US ARMY [3]. During basic training, new recruits have to take on strenuous training protocols including loaded long-distance walking. The lower extremities are therefore exposed to increased and repetitive ground reaction impact forces during basic training. It was found that the most common site of stress fracture in military recruits is the tibia [3], which accounts for more than 40% of total stress fractures in the military [4,5].

Load carriage has been found to alter gait kinematics, and increase ground reaction force proportionally [6,7]. However, the effect of load carriage on the risk of musculo-skeletal injuries such as tibial stress fracture is not fully understood. Specifically, the effect of load carriage on mechanical loading rate has not been investigated. It is not clear, if and how load carriage affects the ground reaction loading rates during walking. The purpose of the study was to investigate the effect of load carriage on ground reaction loading rates during walking. It was hypothesized that ground reaction forces and ground reaction loading rates would increase during loaded walking.

METHODS

Eighteen healthy male subjects (age: 21 ± 2 yr.; body mass: 79 ± 11 kg; body height: 181 ± 4 cm) participated in the study. Subjects wore military boots and performed unloaded walking and loaded walking with a 32 kg rucksack at 1.67 m/s on a

force instrumented treadmill (AMTI). Ground reaction forces were collected at 2400 Hz. The following variables were analyzed: peak vertical and antero-posterior (braking) ground reaction forces, peak instantaneous and average vertical and braking loading rates during weight acceptance of walking. A one-way repeated measures MANOVA was used to determine differences in these variables between normal and loaded walking conditions. $\alpha = 0.05$.

RESULTS AND DISCUSSION

Compared to unloaded walking, the loaded walking exhibited a 49% increase of peak vertical GRF, a 48% increase of peak braking GRF, a 96% increase of vertical ground reaction loading rate, and a 72% increase of braking ground reaction loading rate during weight acceptance ($p < 0.000$) (Table 1).

As we expected, carrying a 32 Kg load led to significant increases of ground reaction forces. As high-magnitude mechanical loads are associated with tibial stress fracture [1, 3, 8], the large increases of ground reaction forces during loaded walking may lead to an increase of tibial bone strain and possibly increase the chance of developing tibial stress fracture.

We also found that there were significant increases of ground reaction loading rates when walking with load. As repeated loading at higher loading rates is more damaging to the bone than at lower loading rates [9], it is possible that the great increase of vertical and braking ground reaction loading rates could expose the tibial bones to increased risk of stress fracture. Surprisingly, the increases of ground reaction loading rates outpace the increases of ground reaction forces during loaded walking. It is possible that it may be the great increase of the ground reaction loading rates leading to great increased risk of tibial stress fracture.

Walking with loads that could be as high as 60% of body weight is an inevitable part of the military basic training. During a twelve-week basic training, the combined running and walking distance could exceed 200 miles [1]. The increased ground reaction forces and ground reaction loading rates associate with every step during loaded walking could expose the military recruits to increased risk of tibial stress fracture. Future study should focus on in-vivo measurement of tibial bone deformation to confirm that load carriage would lead to increases of strain and strain rate during walking.

CONCLUSIONS

Load carriage results in greater increases of ground reaction force loading rates during walking.

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ACKNOWLEDGEMENTS

Funding source: Department of the Army
#W81XWH-08-1-0587

Table 1: Peak vertical and braking GRFs and instantaneous vertical and braking ground reaction loading rates during weight acceptance of walking.

Variables	Unloaded walking	Loaded walking
Peak vertical GRF (BW)*	1.28 ± 0.07	1.90 ± 0.17
Peak braking GRF (BW)*	0.23 ± 0.03	0.34 ± 0.04
Instantaneous vertical ground-reaction loading rate (BW/s)*	17.58 ± 3.96	34.48 ± 10.73
Instantaneous braking ground-reaction loading rate (BW/s)*	8.34 ± 1.73	14.39 ± 5.12

Note. * indicates $p < 0.000$

Appendix D

Abstract presented at the 34th Annual Meeting of American Society of Biomechanics (ASB) in
Provincial, RI, August 18- August 20, 2010

Title: An Integrated Modeling Method for Bone Strain Analysis

AN INTEGRATED MODELING METHOD FOR BONE STRAIN ANALYSIS

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INTRODUCTION

Stress fracture is a common overuse injury experienced by participants of activities involving repetitive loading such as running and loaded walking. One population especially susceptible to this type of injury is US ARMY recruits. The incidence of stress fracture during basic training is about 6% in recruits [1] with the most common site of fracture being the tibia; these tibial stress fractures account for more than 40% of total stress fractures in the military [2,3].

The precise cause of tibial stress fracture remains elusive despite considerable research on the subject [4]. One cause of this ambiguity may be the use of surrogate variables instead of the root mechanical causes of bone microdamage such as deformation. Since gathering bone strain data *in vivo* in large populations has inherent limitations, it is useful to explore musculo-skeletal modeling techniques that can be used to investigate changes in bone strain response across different individuals. The goal of this project was to develop a set of methods capable of generating strain data for large cohorts of subjects.

METHODS

Five subjects (demographics) were used from a larger cohort of subjects whose data were collected for a concurrent project. Walking data were collected for unloaded walking at 1.67 m/s on an AMTI force instrumented treadmill with kinematics collected using 12 Vicon F-series cameras at 120Hz and ground reaction forces collected at 2400Hz (AMTI, Watertown, MA, Vicon, Oxford, UK). Bone geometries were obtained using computed tomography (CT). These images were segmented and 3D geometry files were generated in Materialise MIMICS 13.0 (Materialise, Leuven, Belgium).

These 3D surface geometries were used in MD MARC 2008 (MSC.Software, Santa Anna, CA) to build hexmesh finite element (FE) models with generic linear isotropic material properties of elastic modulus 17GPa, density 1.9g/cm³, and Poisson's ratio of 0.3. These tibia models were then imported into a scaled LifeMOD musculo-skeletal model.

Once positioned, spatial coordinates of muscle model markers representing origins and insertions as well as joint positions relative to the tibia were exported. The exported position data were used to build new FE tibias incorporating massless rigid body links between muscle and joint attachment points using a custom Python script (version 3.0). Boundary conditions were assigned as rotational and translational degrees of freedom of the nodes representing the ankle and knee joint centers and flexible bodies (modal neutral files) were generated for use in LifeMOD.

For a full description of LifeMOD modeling methods and the use of flexible bodies, see Al Nazer [5]. Key differences between the current study and that of Al Nazer et al include a larger number of muscle actuators on the leg that includes the flexible tibia, subject specific segment scaling based on joint center calculations, ground reaction forces, and subject specific tibial geometries generated from CT scans.

In order to build a muscle and joint actuated subject specific model, a static lower body model was first built using subject sex, mass, and height as scaling inputs into the GeBOD [6] database incorporated into LifeMOD. Segments and joint locations and orientations were then scaled using joint center data calculated in Visual3D. Once a scaled model was built, experimental kinematic data were then used to perform an inverse

kinematic (IK) analysis. The results of this analysis were used to “train” the muscle (right leg) and joint (left leg) PID controllers used to actuate the model in a forward dynamics (FD) analysis. Flexible tibias were then imported into LifeMOD.

A FD analysis was then performed with the addition of ground reaction forces applied to both feet and motion capture kinematics disabled. Maximum principle, minimum principle, and maximum shear strain values were then calculated using the Durability plug-in for MD ADAMS/View for the nodes most closely representing the location of the strain staples used in *in vivo* studies [5,7,8,9] (MSC. Software, Santa Ana, CA).

RESULTS AND DISCUSSION

The averaged results for a preliminary analysis of 5 subjects are presented in Table 1. Maximum principle and minimum principle strains and strain rates appear to be consistent with previously reported values, though maximum shear strain and strain rate values for these 5 subjects are markedly lower. Further analysis is needed to determine if these differences are due to modeling methods or to differences in experimental protocols, specifically treadmill versus over-ground walking.

In summary, the methods used in this study provide a promising means of investigating tibial strain and strain rate across individuals and

conditions. Since these methods do not require the use of invasive bone staples, more diverse populations and conditions can be studied in the future.

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ACKNOWLEDGEMENTS

Funding source: Department of the Army
#W81XWH-08-1-0587

Ed Rezer of Device Analytics for FE analysis support

Table 1: Averaged Strains and Strain Rate Comparisons

	Strain (Microstrain)			Strain Rate (Microstrain/s)		
	Max Prin	Min Prin	Max Shear	Max Prin Rate	Min Prin Rate	Max Shear Rate
Lanyon et al ⁷	395	-434	829	Not reported	-4000	Not reported
Burr et al ⁸	437	-544	871	11006	-7183	16162
Milgrom et al ⁹	840	-454	1183	3955	-3306	10303
Al Nazer et al ⁵	305	-645	948	4000	-7000	10000
Present Simulation	526	-531	426	5042	-2548	3062

Appendix E

Abstract presented at the 58th annual meeting of American College of Sports Medicine (ACSM) in Denver, CO. June 1- June 4, 2011.

The abstract was published in *Medicine & Science in Sports & Exercise*. V. 43, No. 5 Supplement, S21-22

Title: Effects of Load Carriage and Muscular Fatigue on Ground Reaction Loading Rate during Walking

Medicine & Science in Sports & Exercise:

May 2011 - Volume 43 - Issue 5 - p 30

doi: 10.1249/01.MSS.0000402761.29699.81

C-15 Thematic Poster - Obesity and Gait: JUNE 2, 2011 8:00 AM - 10:00 AM: ROOM: 101

Effects Of Load Carriage And Muscular Fatigue On Ground Reaction Loading Rate During Walking: 629: Board #4 8:00 AM - 10:00 AM

Wang, Henry¹; Frame, Jeff¹; Ozimek, Elicia¹; Reedstrom, Cara²; Leib, Daniel²; Dugan, Eric²

¹Ball State University, Muncie, IN. ²Boise State University, Boise, ID.

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(*No relationships reported*)

Military personnel are often afflicted by tibia stress fracture (TSF) during basic training. Load carriage and muscle fatigue may be factors related to the high incidence of TSF. Load carriage increases ground reaction force during walking. Muscle fatigue reduces muscles' ability to attenuate dynamic load on musculoskeletal system during locomotion. However, it is yet to be determined what effects load and fatigue may elicit in mechanical loading rate during walking.

PURPOSE: To determine the effects of load carriage and fatigue on peak vertical ground reaction force and loading rate during walking.

METHODS: Eighteen healthy males (age: 21 ± 2 yrs, body mass: 77.6 ± 9.6 kg, body height: 181 ± 4 cm) performed tasks in the following order: unloaded and unfatigued walking (UU), loaded and unfatigued walking with a 32 kg rucksack (LU), a fatiguing protocol consisting of stepping and heel-raising with a 16 kg rucksack, loaded and fatigued walking with a 32 kg rucksack (LF), and unloaded fatigued walking (UF). Each walking task was performed for 5 min on a force instrumented treadmill (AMTI) at 1.67 m/s. Peak vertical ground reaction force and loading rate at weight acceptance were normalized to body weight. One way repeated measures ANOVAs and pair-wise comparisons with Bonferroni correction were performed. $\alpha = 0.05$.

RESULTS: The peak vertical ground reaction forces of the UF (1.35 ± 0.11 BW), LU (1.92 ± 0.18 BW), and LF (1.99 ± 0.19 BW) were all greater than that of the UU (1.27 ± 0.06 BW) ($p < 0.05$). The peak vertical ground reaction loading rates of the UF (21.75 ± 7.92 BW/s), LU (35.29 ± 12.07 BW/s), and LF (37.58 ± 11.92 BW/s) were all greater than that of the UU (16.81 ± 3.40 BW/s) ($p < 0.05$).

CONCLUSION: Fatigue, load carriage, and the combination of load carriage and fatigue all lead to significant increases of mechanical loading and loading rate on the lower extremities during walking. As great magnitudes of mechanical loading and loading rate are identified risk factors of TSF, the effects of load carriage and muscle fatigue could expose military personnel to increased risk of TSF during basic training. Interestingly, the increases of mechanical loading rate are more pronounced than that of the mechanical loading. Mechanical loading rate may have greater effect on the development of TSF than the loading magnitude. DOD#W81XWH-08-1-0587

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Appendix F

Abstract presented at the 58th annual meeting of American College of Sports Medicine (ACSM) in Denver, CO. June 1- June 4, 2011.

The abstract was published in *Medicine & Science in Sports & Exercise*. V. 43, No. 5 Supplement, S639

Title: The Effect of Height on Tibial Strain during Drop Landings

Medicine & Science in Sports & Exercise:
May 2011 - Volume 43 - Issue 5 - pp 918-919
doi: 10.1249/01.MSS.0000402566.81347.54

G-39 Free Communication/Poster - Modeling: JUNE 4, 2011 7:30 AM - 11:00 AM: ROOM: Hall B

The Effect of Height on Tibial Strain During Drop Landings: 3174: Board #137 June 4 9:30 AM - 11:00 AM

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(No relationships reported)

During landing, the human body is required to absorb impact forces throughout its tissues. Muscle and connective tissue dissipate much of this force. A portion of the impact is delivered to the bones. Repetitive forces acting on the tibia during landing can cause microfractures which may lead to stress fracture. It is difficult to quantify bone strain through invasive measure such as bone staple.

PURPOSE: To calculate changes in bone strain during landing using a noninvasive approach by integrating a finite element tibia in a musculoskeletal model.

METHODS: A musculoskeletal model representing a healthy male subject (22 years, 78.6 kg, 1.85 m) was created. A flexible tibia, created from a CT scan of the subject's right tibia, was included in the model. Motion capture data were collected while the subject performed drop landings from three separate heights (26, 39, and 52 cm) and served as inputs to perform dynamic simulations. Surface electromyography and joint angle data were compared to their simulation using a cross correlation. Maximum magnitudes of principal and maximum shear strain were computed.

RESULTS: Strong cross-correlation coefficients ($CCC > 0.87$) were found between recorded and simulated lower extremity joint angles. Recorded activity in the vastus lateralis, vastus medialis, and the tibialis anterior agreed strongly with the model ($CCC > 0.75$). Weaker relationships existed in the gastrocnemius, hamstring, and soleus, which may be due to co-contraction. Maximum principal strain increased with height in 26cm ($595 \pm 136 \mu\text{strain}$), 39cm ($879 \pm 134 \mu\text{strain}$), and 52cm ($1077 \pm 108 \mu\text{strain}$). Minimum principal strain negatively increased with height in 26cm ($-593 \pm 86 \mu\text{strain}$), 39cm ($-645 \pm 35 \mu\text{strain}$), and 52cm ($-769 \pm 33 \mu\text{strain}$) landings. Maximum shear strain followed the same increasing trend in 26cm ($592 \pm 111 \mu\text{strain}$), 39cm ($760 \pm 83 \mu\text{strain}$), and 52cm ($899 \pm 98 \mu\text{strain}$) landings.

CONCLUSION: The simulated results showed reasonable agreement with the recorded lower extremity joint angles and muscle activities. The calculated tibial strains were in reasonable agreement with the literature. This study demonstrates a valid non-invasive method of simulating landing movement and computing tibial strain produced during landing movements.

Supported by DoD #W81XWH-08-1-0587 and BSU 2009 ASPiRE.

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Appendix G

Abstract presented at the 35th Annual Meeting of American Society of Biomechanics (ASB) in Long Beach, CA, August 10- August 14, 2011

Title: Influences of Load Carriage and Fatigue on Lower-extremity Kinetics during Walking

INFLUENCES OF LOAD CARRIAGE AND FATIGUE ON LOWER EXTREMITY KINETICS DURING WALKING

¹ He Wang, ¹Jeff Frame, ¹Elicia Ozimek, ²Cara Reedstrom, ²Daniel Leib, and ²Eric Dugan,
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INTRODUCTION

Military personnel are commonly afflicted by lower extremity overuse injuries such as knee pain and stress fractures [1, 2]. Walking with heavy loads is an inevitable part of the military training and during the twelve-weeks of basic training, the combined running and walking distance could exceed 200 miles [1]. Therefore, military personnel commonly face physical challenges comprised of load carriage and muscle fatigue.

Load carriage has been found to alter gait kinematics [3]. Specifically, there are increases of pelvic anterior tilt, hip flexion, and knee flexion angles at heel strike [3, 4]. Ground reaction forces and ground reaction loading rates are also increased during loaded walking [3, 5]. Muscle fatigue has also been found to alter running kinematics with increases of hip and knee angles at heel strike [6, 7]. During fatigued walking, vertical ground reaction force and loading rate are found to increase [8]. Further, fatigued muscles' ability to attenuate impact loading is decreased [9].

Thus, under the influences of load carriage and muscle fatigue, the lower extremities are exposed to increased ground reaction impact forces with increased loading rate during walking. The risk of lower extremity injury may be increased. However, it is yet to be determined if the lower extremity joint kinetics are altered as a result of load carriage and muscle fatigue. Analyzing lower extremity joint kinetics during loaded and fatigued walking will broaden our knowledge on the potential causes for development of lower limb injuries during military training.

The purpose of the study was to investigate the lower extremity joint kinetics during loaded and fatigued walking. As the vertical ground reaction force and loading rate are increased during weight

acceptance, it is expected that the lower extremity joint kinetics will be altered to accommodate the increased external impact loading.

METHODS

Eighteen healthy male subjects (age: 21 ± 2 yr.; body mass: 79 ± 11 kg; body height: 181 ± 4 cm) participated in the study. Subjects wore military boots and participated in a fatiguing protocol which involved a series of metered step-ups and heel raises while wearing a 16 kg rucksack. Subjects performed the following tasks in sequence: 5-min unloaded walking; 5-min loaded walking with a 32 kg rucksack; Fatiguing protocol; 5-min loaded walking with a 32 kg rucksack under fatigue; 5-min unloaded walking under fatigue. All walking tasks were performed at 1.67 m/s on a force instrumented treadmill (AMTI). A 15-camera system (VICON) was used to track reflective markers placed on the human body at 120 Hz. Ground reaction forces were collected at 2400 Hz. Visual 3D (C-Motion) was used to calculate lower extremity joint kinetics. The following variables were analyzed: peak knee and hip extensor moments and peak knee and hip joint powers during weight acceptance of walking. Two-way repeated measures ANOVAs were performed. Load carriage and fatigue were the independent factors. $\alpha = 0.05$.

RESULTS AND DISCUSSION

No interactions were found between load carriage and fatigue ($P > 0.05$). Load carriage led to significant increases of knee and hip extensor moments, knee and hip joint powers ($P < 0.001$) (Table 1). Fatigue did not lead to changes in knee extensor moment and knee joint power ($P > 0.05$) but resulted in significant increases of hip extensor moment and hip joint power ($P < 0.01$) (Table 1).

In this study, we found the knee extensor moment increases as a partial mechanism to absorb the increased ground impact forces during loaded walking. Therefore, during long-distance loaded walking, it is possible that the increased knee power absorption along with high magnitudes of impact forces could expose the military recruits to increases of overuse knee injuries. We also found the hip joint exhibited increased extensor moment and joint power magnitude during weight acceptance. As increased pelvic anterior tilt is associated with load carriage [4], the increased hip extensor moment may be used to decelerate the increased pelvic anterior tilt at heel strike.

During fatigued walking, the hip extensor moment and power increase during weight acceptance. The increased hip extensor moment may be used to stabilize the pelvis and elevate the center of mass. Also, the increased hip extensor moment can help stabilize the femur and prevent knee flexion during weight acceptance. Interestingly, Fatigue does not lead to alteration of knee extensor moment and knee power absorption during weight acceptance. Thus, the increased vertical ground reaction force during fatigued walking may be attenuated by lower leg bone structures such as the tibia.

In summary, during weight acceptance, load carriage leads to alterations of knee and hip joint

kinetics; Fatigue leads to alterations of hip kinematics. The altered lower extremity joint kinematics associated with load carriage and fatigue may be related to the high incidence of lower extremity overuse injuries in the military.

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ACKNOWLEDGEMENTS

Funding source: US ARMY #W81XWH-08-1-0587

Table 1: Means and SDs of Peak knee and hip extensor moments and peak knee and hip joint powers during weight acceptance of walking.

Variables	Unloaded and Unfatigued	Loaded and Unfatigued	Loaded and Fatigued	Unloaded and Fatigued
Knee extensor moment (Nm/kg)*	0.88 (0.20)	1.61 (0.37)	1.63 (0.42)	0.90 (0.25)
Hip extensor moment (Nm/kg)* #	1.54 (0.41)	2.26 (0.42)	2.38 (0.42)	1.85 (0.48)
Knee joint power (W/kg)*	-1.34 (0.37)	-2.65 (1.40)	-2.76 (1.06)	-1.39 (0.54)
Hip joint power (W/kg)* #	1.03 (0.36)	1.69 (0.61)	1.97 (0.64)	1.47 (0.49)

Note. * indicates significant difference between loaded and unloaded walking conditions ($P < 0.00$); # indicates significant difference between fatigued and unfatigued walking conditions ($P < 0.01$).

Appendix H

Abstract presented at the 35th Annual Meeting of American Society of Biomechanics (ASB) in Long Beach, CA, August 10- August 14, 2011

Title: A Time-efficient Method for Analyzing Bone Strain with Large Subject Pools

A TIME-EFFICIENT METHOD FOR ANALYZING BONE STRAIN WITH LARGE SUBJECT POOLS

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INTRODUCTION

Bone strain is a useful measurement when investigating overuse injuries such as stress fracture [1]. *In-vivo* data for bone strain is difficult to obtain due to the invasiveness of surgical procedures and even when such studies can be conducted, interpretation of data from implanted strain staples and gauges is non-trivial [2].

Due to the difficulties in measuring *in-vivo* strain there has been some momentum in utilizing numerical modeling methods to investigate strain. Finite element modeling (FEM) has been used in past investigations in conjunction with mechanical testing of cadaver specimens to investigate certain aspects of bone stress and strain. This methodology is quite time consuming computationally, however, and can be difficult to use in conjunction with subject-specific kinematic and kinetic data, thus limiting its application in investigations requiring large subject pools.

Recent work has been done to overcome the computational time and de-coupling of FEM inputs by Al Nazer and Kłodowski using a flexible body in conjunction with a subject-specific musculoskeletal model [3]. The data and methods presented in these papers still suffer from limitations of small sample size and using simulation-generated ground reaction forces, however, and do not present methods to manage large sets of data.

It is the purpose of this paper to present a time-efficient methodology of calculating bone strain utilizing subject-specific tibial geometry and material properties derived from computed tomography (CT) scans and musculoskeletal models driven by experimental kinematic and kinetic inputs. Data and computational times from a single

subject and gait trial will be presented as an example of how this methodology can be used with implications for automation across large cohorts.

METHODS

A representative subject from a larger study on tibial stress fracture was used to describe this process. All experimental procedures were approved by the University's Institutional Review Board and the subject signed an informed consent form. The subject was a 21 year old male.

CT scans were collected for the entire length of the tibia using a GE Light Speed VCT (General Electric, USA). Images were segmented in Mimics 12.1 (Materialise, Belgium), a surface mesh automatically generated, and a 3mm³ solid hexahedral mesh generated from that surface mesh using MD MARC 2008 (MSC.Software, Santa Ana, CA). An automated custom MATLAB routine was used to assign material density and elastic modulus to each individual element of the mesh based on the average Hounsfield unit value of the pixels contained within each element; this process was automated across the full cohort for the larger study as well. Six hundred material properties were used in this assignment with 300 each being considered cortical bone [4] and 300 cancellous [5].

Motion capture data were collected using a cluster-based marker set at 120Hz while the subject walked at 1.67m/s on an in-line AMTI force instrumented treadmill collecting analog data at 2400Hz.

A scaled lower-body model was built using LifeMOD 2008.2805 (Lifemode, San Clemente, CA) based on the subject characteristics [6]. Twenty three muscles were then added to the right leg [7]. The tibia constructed in MARC was then exported

and manually aligned in the musculoskeletal model using digitized landmarks. Relative locations of relevant attachment sites were then exported and applied to the FEM tibia in MARC using an automated Python script and a Craig-Bampton modal analysis performed with 6 degrees of freedom applied to each "joint" node. This process was automated using the MD ADAMS command language and references to text files containing subject characteristics and tibia node numbers.

An inverse-kinematic (IK) trial was performed using experimental trajectory data to set kinematic goals for muscle and joint controllers and then the modal neutral file imported to align with three known points in the musculoskeletal model. The rationale for using an MNF is described more fully in [8]. A forward dynamic simulation was then performed using inverse-kinematic results as targets and experimental ground reaction forces applied. Strains were then calculated using the Durability plug-in for MD ADAMS/View (MSC.Software, Santa Ana, CA). This IK through strain calculation process was automated using the MD ADAMS command language and automated across the larger cohort using MATLAB and the Windows command line.

RESULTS AND DISCUSSION

Strain and strain rate data generated by the model fall well within expected values for gait. An example stride of strain data is shown in Figure 1 and results summarized in Table 1.

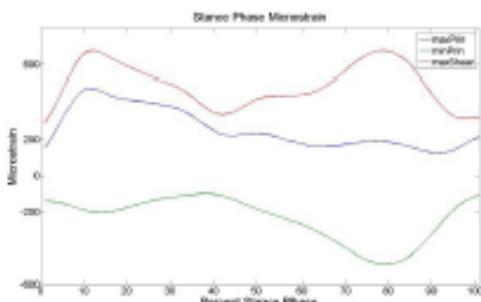


Figure 1: A representative stance phase of strain data.

	Strain Magnitude (Microstrain)			Strain Rate (Microstrain/s)		
	Max Prin	Min Prin	Max Shear	Max P Rate	Min P Rate	Max Shear Rate
Lanyon et al	395	-434	829	Not reported	-4000	Not reported
Burr et al	437	-544	871	11006	-7183	16162
Milgrom et al	840	-454	1183	3955	-3306	10303
Al Nazer et al	305	-645	948	4000	-7000	10000
Present Simulation	490	-515	725	6608	-2804	7305

Table 1: Summary of strain results compared to previous studies

The methods utilized in this study are time efficient and highly automatable. User interaction is required through software only to segment CT images (though this is largely automated), select nodes for which to compute strain data, and to orient the tibia model within the musculoskeletal model. These tasks collectively take less than one hour per subject for an experienced operator. Simulation times for this method were easily manageable for large groups, taking only 5.5 minutes per 5 seconds of gait. Calculation of strains in Durability took only 0.5 seconds per node for a 5 second gait trial. With well-planned software automation utilizing scripting capabilities of MARC, LifeMOD, and MATLAB, large volumes of data can be handled reasonably for strain studies requiring large sample sizes.

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Appendix I

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Title: Influence of Fatigue and Load Carriage on Mechanical Loading during Walking

Influence of Fatigue and Load Carriage on Mechanical Loading During Walking

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ABSTRACT Load carriage and muscular fatigue are two major stressors experienced by military recruits during basic training. The purpose of this study was to assess the influences of load carriage and muscular fatigue on ground reaction forces and ground reaction loading rates during walking. Eighteen healthy males performed the following tasks in order: unloaded and unfatigued walking, loaded and unfatigued walking, fatiguing exercise, loaded and fatigued walking, and unloaded and fatigued walking. The fatiguing exercise consisted of a series of metered step-ups and heel raises with a 16-kg rucksack. Loaded walking tasks were performed with a 32-kg rucksack. Two-way repeated measures analysis of variances were used to determine the effects of fatigue and load carriage on ground reaction forces and loading rates. Muscular fatigue has a significant influence on peak vertical ground reaction force and loading rate ($p < 0.01$). Load carriage has a significant influence on peak ground reaction forces and loading rates ($p < 0.001$). As both muscular fatigue and load carriage lead to large increases of ground reaction forces and loading rates, the high incidence of lower extremity overuse injuries in the military may be associated with muscular fatigue and load carriage.

INTRODUCTION

Military personnel are commonly afflicted by lower extremity overuse injuries.^{1–3} Major overuse injuries such as stress fracture could result in significant medical cost and loss of training days to the military organizations.⁴ The high incidence of lower extremity overuse injuries in the military is associated with high volume and high intensity of physical training.^{1–3} Loaded long-distance running and walking are a major part of the strenuous training protocol during basic training. The combined mileage of walking and running during a 12-week basic training could exceed 200 miles.¹ Mechanical impact loadings from the ground such as the ground reaction force and the loading rate are introduced during heel strikes of walking/running. The high volume of repetitive ground impact forces experienced by military personnel may contribute to increased risk of lower extremity overuse injuries. In fact, it was reported that high magnitude ground reaction forces and loading rates could lead to increased risk of lower extremity overuse injuries.^{1,5–7}

Common overuse injuries occurring during basic training include knee pain, back pain, and stress fracture.³ Stress fractures are among the severe lower extremity musculoskeletal injuries demanding extended periods of recovery.^{5,8} The injury rate of stress fractures during basic training is approximately 6% in the U.S. Army⁵ and could be as high as 31% in the Israeli Army.⁹ Stress fracture is a result of excessively repetitive loads acting on the bone, leading to fatigue-induced bone microdamage.^{1,10} It has been proposed that when loading accumulates at a faster rate than the bone's remodeling process, microdamage accumulates and micro-

scopic cracks are formed and propagated in the bone.⁵ Bone strength is weakened during this stage, and a stress fracture may eventually occur.¹ Loading rate is an important mechanical factor related to stress fracture development.^{1,5,11} Increased ground reaction loading rate has been linked to tibial stress fracture in female runners.^{12,6} High-magnitude loads are also found to be associated with stress fracture,^{1,5} and runners with a history of stress fracture tend to exhibit greater ground reaction forces.⁷ Thus, ground reaction force and ground reaction loading rate may be two important mechanical factors related to tibial stress fracture.

Skeletal muscles play an important role in the attenuation of loads applied to the skeletal system. During weight acceptance in activities such as running and drop landing, leg muscles contract eccentrically to attenuate ground reaction impact forces.¹³ Muscular fatigue decreases muscle force generation and can reduce muscle's ability to attenuate ground reaction impact forces.^{14–16} Although it is possible that muscular fatigue may expose lower extremity to larger mechanical loads during impact-related activities, few studies have been conducted to support this hypothesis. Information in the literature is limited and often times contradictory. For example, it was reported that muscular fatigue resulted in an increased vertical ground reaction force and ground reaction loading rate during landing¹⁷ and running.¹⁸ However, it was also found that muscular fatigue could lead to a decreased vertical ground reaction force and loading rate during running.¹⁹ Therefore, it is necessary to continue to investigate if and how muscular fatigue affects ground reaction forces during activities involving repetitive ground impact loading such as walking.

Walking with load carriage has been investigated in the past.^{20–22} Load carriage results in alterations of gait kinematics such as reduced stride length, increased cadence, and increased pelvic tilt and hip flexion.^{23,24} It was also found that walking with additional load accompanies with an

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increased vertical ground reaction force.^{20–22} The increase of vertical ground reaction force was proportional to the amount of load carried.^{21,23,25} Although it is clear that heavy load carriage results in increased ground reaction forces, the influence of load carriage on ground reaction loading rates has yet to be determined.

Further, the combined effect of fatigue and load carriage on ground reaction forces needs to be examined. It is not known if there is an interaction between these two factors. Determining the characteristics of ground reaction forces when both effects of fatigue and load carriage are presented will expand our understanding on mechanisms of lower extremity overuse injuries.

During basic training, the strenuous training protocol exposes military recruits to significant physical fatigue.²⁶ In addition, walking with heavy load is always an important part of the training.²⁷ Prolonged walking with load carriage could lead to significant neuromuscular impairment.²⁶ Thus, fatigue and load carriage are two major stressors experienced by military recruits. Given the high incidence of overuse injuries during basic training and the injury mechanisms are far from understanding, it is warranted to investigate the effects of fatigue and load carriage on ground reaction forces. It is possible that the fatigue and load carriage may have negative effects on ground reaction forces during walking.

Therefore, the purpose of this study was to investigate the influences of fatigue and load carriage on ground reaction forces and ground reaction loading rates during walking. We hypothesized that there would be an increased vertical ground reaction force and increased ground reaction loading rates during walking in a fatigued state. Also, there would be increased ground reaction forces and ground reaction loading rates during walking with load carriage.

METHODS

This study is a part of a larger, ongoing research project investigating the effects of fatigue and load carriage on walking biomechanics. Eighteen healthy college male participants were recruited to the study. The means and standard deviations (SDs) of age, body mass, body height, and $\dot{V}O_2$ max of participants were 21(2) years, 77.6(9.6) kg, 181(4) cm, and 51.4(5.3) mL kg^{-1} min^{-1} , respectively. Participants were recreationally active, classified as low risk for cardiovascular diseases according to American College of Sports Medicine guidelines,²⁸ and free from known musculoskeletal injury. All participants met the military enlistment standards in terms of physical conditions.²⁹ In addition, the age, body mass, body height, and fitness level ($\dot{V}O_2$ max) of our participants were comparable to those military recruits entering the U.S. Army basic training.³⁰ Institutional Research Board approval was obtained before commencing the study.

A tandem force instrumented treadmill (AMTI Advanced Mechanical Technology, Watertown, Massachusetts) with two force platforms installed under the tandem belts was used to control the walking speed at 1.67 m/s, while allowing the

ground reaction forces to be detected. Reflective markers were attached on both sides of the body in the following locations to track the walking movement: acromion, sternum, anterior superior iliac spine, posterior superior iliac spine, lateral knee, lateral ankle, heel, base of the fifth metatarsal, and base of the second toe. In addition, two cluster marker sets were attached on the thigh and shank, respectively. Fifteen VICON MX and F-20s series cameras (VICON, Denver, Colorado) were used to track the reflective markers in the space. VICON Nexus (V 1.4.116) was used to collect kinematic data at 120 Hz and ground reaction forces at 2,400 Hz.

Participants wore compression shorts, a compression shirt, and military boots (Altama Mil Spec Desert 3 Layer boot) during the experiment. Participants walked at a self-selected pace on the force-instrumented treadmill for 5 minutes to warm up. After the warm-up, participants' maximal vertical jump heights were assessed by using a Vertec (Sports Imports, Columbus, Ohio). Three attempts were made and the highest of the three was used to determine the 80% of the jump height. The general experimental protocol consisted of tasks in the following order: (a) 5-minute normal walking in an unloaded and unfatigued state (UU); (b) 5-minute walking in a loaded and unfatigued (LU) state with a 32-kg rucksack (Modular Lightweight Load-Carrying Equipment; Specialty Defense System, Dunmore, Pennsylvania); (c) fatiguing protocol; (d) 5-minute walking in a loaded and fatigued state (LF) with a 32-kg rucksack; (e) 5-minute walking in an unloaded and fatigued state (UF). The 32-kg load carriage used in this study represented a typical approach/marching load experienced by the military personnel.³¹ The fatiguing protocol was carried out after the completion of the loaded walking task and immediately followed by the LF walking task. No rest time was given between the tasks. The treadmill speed was set at 1.67 m/s for all participants in all conditions. Ten trials were collected during the 5-minute interval of each walking condition.

The goal of the fatiguing protocol was to induce a significant level of muscular fatigue in the lower extremities, defined as a decline of muscle force output in response to voluntary effort.³² Therefore, participants completed circuits that included loaded stepping and heel raises. The stepping protocol was based on a modified Queens College Step Test procedure;³ specifically, the participant performed the test while wearing a 16-kg rucksack (Modular Lightweight Load-Carrying Equipment, Specialty Defense System), the step height was set at 16 inches, and the participant stepped up and down at a rate of 24 cycles per minute. One cycle was defined as step up with first leg, step up with second leg, step down with first leg, and step down with second leg. The first leg was alternated to ensure even loads across both legs. The participant performed this stepping sequence until he could no longer match the cadence of the metronome. At this time, the participant completed 20 heel raises standing at the edge of a box.

Following the heel raises, the participant removed the rucksack and completed a maximal effort vertical jump using

the Vertec to quantify the level of fatigue. This sequence was repeated until the participant's maximal vertical jump fell below 80% of the maximal jump height. Researchers provided verbal encouragement throughout the protocol in an effort to elicit maximal effort from the participant.

Experimental data were processed in Visual 3Dv4.0 (C-Motion, Germantown, Maryland). The following dependent variables were analyzed: peak vertical ground reaction force, peak anteroposterior (braking) ground reaction force, peak vertical ground reaction loading rates, and peak anteroposterior (braking) ground reaction loading rates during weight acceptance of the stance. Ground reaction forces were filtered using a zero-lag Butterworth filter with a cutoff frequency of 40 Hz. Ground reaction loading rates were calculated between 20% and 80% of the period between heel strike and peak ground reaction forces (vertical and anteroposterior) during weight acceptance of walking.⁶ Peak ground reaction loading rate was the peak loading rate occurring during the period. In addition, the ground reaction forces and ground reaction loading rates were normalized to body weight.

SPSS v16 (SPSS, Chicago, Illinois) was used to perform statistical analysis. Two-way repeated measures analysis of variance was used to determine the effects of fatigue and load carriage. Bonferroni correction was used to perform the pairwise comparisons for dependent variables. Significance level was set at 0.05.

RESULTS

No interaction was found between the effects of load carriage and fatigue for peak vertical ground reaction force ($p = 0.589$), peak braking ground reaction force ($p = 0.239$), peak vertical ground reaction loading rate ($p = 0.240$), and peak braking ground reaction loading rate ($p = 0.307$). The fatigue had a significant effect on peak vertical ground reaction force ($p < 0.001$). The load carriage had a significant effect on both peak vertical and braking ground reaction forces ($p < 0.001$). Furthermore, the fatigue had a significant effect on peak vertical ground reaction loading rate ($p = 0.003$) and a near significant effect on peak braking ground reaction loading rate ($p = 0.084$). The load carriage had a significant effect on both the peak vertical and braking ground reaction loading rates ($p < 0.001$).

The means and SDs of the mechanical loading variables at weight acceptance during the four walking conditions are

presented in Table I. The UF, LU, and LF conditions showed large increase of peak vertical ground reaction force (6%, 51%, and 57%, respectively) than the UU condition ($p = 0.002$, $p < 0.001$, $p < 0.001$, respectively). Both the LU and LF conditions showed large increase of peak braking ground reaction force (52% and 48%, respectively) than the UU condition ($p < 0.001$). Furthermore, the peak vertical ground reaction loading rates of the UF, LU, and LF conditions were 29% ($p = 0.045$), 110% ($p < 0.001$), and 124% ($p < 0.001$), which is greater than that of the UU condition, respectively. The peak braking ground reaction loading rates of the UF, LU, and LF conditions were 23% ($p = 0.007$), 86% ($p < 0.001$); and 93% ($p < 0.001$), which is greater than that of the UU condition, respectively.

DISCUSSION

The purpose of the study was to assess the effects of muscular fatigue and load carriage on ground reaction forces and ground reaction loading rates during walking. In this study, muscular fatigue was introduced through a fatiguing protocol consisting of a series of stepping exercise and heel raise exercise. A 32-kg rucksack was added to the body to represent the load carried. Participants performed the following four walking tasks: UU walking, LU walking, LF walking, and UF walking. The influences of the muscular fatigue and load carriage on ground reaction forces and loading rates were then assessed.

During the fatiguing exercise, participants reached a fatigued state if they were not able to jump to 80% of their maximal prefatigue jump heights. It is known that muscular fatigue decreases muscle force generation and can reduce muscle's ability to attenuate ground reaction impact forces.^{14–16} Thus, we hypothesized that there would be an increase of peak vertical ground reaction force during weight acceptance of walking in a fatigued state. Our hypothesis was supported. In this study, we found that leg muscle fatigue had a significant effect on peak vertical ground reaction force. The effect of muscle fatigue on walking movement was not documented well in the past. However, studies investigating running and drop-landing types of activities have reported that muscle fatigue led to increase of ground impact forces.^{17,18} Our findings are in agreement with these studies and confirmed that walking in a fatigued state is accompanied by increased vertical ground reaction force. Specifically,

TABLE I. Means and SDs of Peak Vertical and Braking Ground Reaction Forces and Peak Vertical and Braking Ground Reaction Loading Rates of the UU, UF, LU, and LF Conditions During Weight Acceptance of Walking

	UU	UF	LU	LF
Peak Vertical Ground Reaction Force (BW)	1.27 (0.06)	1.35 (0.11) ^a	1.92 (0.18) ^a	1.99 (0.19) ^a
Peak Braking Ground Reaction Force (BW)	0.23 (0.03)	0.24 (0.03)	0.35 (0.06) ^a	0.34 (0.06) ^a
Peak Vertical Ground Reaction Loading Rate (BW/s)	16.81 (3.40)	21.75 (7.92) ^a	35.29 (12.07) ^a	37.58 (11.92) ^a
Peak Braking Ground Reaction Loading Rate (BW/s)	7.98 (1.61)	9.78 (2.06) ^a	14.81 (5.89) ^a	15.44 (4.36) ^a

^aIndicates significant difference from the UU condition ($p < 0.05$).

there was a 6% increase of peak vertical ground reaction force at weight acceptance during UF walking.

We further hypothesized that the leg muscle fatigue influences the peak ground reaction loading rates. This hypothesis was supported. We found that leg muscle fatigue had a significant effect on the peak vertical ground reaction loading rate. In addition, leg muscle fatigue showed a tendency to affect the peak braking ground reaction loading rate ($p = 0.084$). In particular, compared to UU walking, during the UF walking, there were 29% and 23% increase in peak vertical and braking ground reaction loading rates, respectively. Our results are in agreement with previous studies,^{17,18} which reported that fatigue led to increase of vertical ground reaction loading rate during running and drop-landing activities.

Although there have not been many studies investigating the fatigue effect on ground reaction forces and loading rates during impact-related activities, the majority of the studies reported increase of ground impact forces and loading rates during activities such as running and drop landing.^{8–18} However, contradictory results have also been reported.^{8,19} In the study conducted by Gerlach et al,¹⁹ decreased ground impact force and loading rates were shown. The changes in ground impact forces were thought to be the results of altered running cadence, step length, and lower extremity joint kinematics.¹⁹ It is worth noting that runners recruited in Gerlach's study were experienced runners.¹⁹ It is possible that experienced runners know how to adjust their running kinematics to reduce the chance of increasing ground impact forces associated with muscle fatigue. In our study, we recruited a group of college students who had no experience of military basic training as well as no experience of walking under the influence of fatigue. Thus, findings from our study imply that when participating in unfamiliar training programs consisting of fatigued walking, minimally conditioned trainees may experience increased vertical ground reaction force and loading rate.

In this study, we also assessed the effect of load carriage on ground reaction forces. We hypothesized that walking with a 32-kg rucksack would result in the increase of ground reaction forces. Our hypothesis was supported. We found that load carriage had a significant effect on ground reaction forces. Specifically, there were 50% increase of the peak vertical and braking ground reaction forces at weight acceptance during the LU walking when compared to the UU walking. Results from this study are also consistent with the findings from previous studies.^{23,24}

Although the characteristics of ground reaction loading rates during loaded walking have not been analyzed in the past, we did expect that the load carriage would lead to increased ground reaction loading rates at weight acceptance of walking. Not surprisingly, we found that load carriage led to significant increase of peak ground reaction loading rates during walking. In particular, when compared to UU walking, the LU walking exhibited a 110% increase of peak vertical ground reaction loading rate and an 86% increase of peak braking ground reaction loading rate.

In this study, there was no interaction between the fatigue and load carriage effects for the ground reaction forces and loading rates. Similar to the effects of fatigue and load carriage, the combined effect of fatigue and load carriage led to significant increase of ground reaction forces and loading rates. Compared to UU walking, there were 57% and 48% increase of vertical ground reaction force and braking ground reaction force associated with the LF walking, respectively. The increase of ground reaction loading rates were even more pronounced with the loading rate more than doubled (124%) in the vertical direction and almost doubled (93%) in the anteroposterior direction. Therefore, lower extremities were exposed to great mechanical loads at fast rates during the LF walking.

High vertical ground reaction force and loading rate are biomechanical risk factors for lower extremity overuse injuries.^{1,5–7} Thus, training under the influences of fatigue and load carriage could increase the risk of developing overuse injuries. Among the common overuse injuries experienced by military recruits, stress fracture has been identified as severe and leads to significant loss of training days.^{1,5} Tibial stress fracture contributes to a majority of the stress fracture reported in the past.^{34,35} As increased mechanical load and loading rate lead to increase of bone strain and strain rate,^{1,5,36} heavy load carriage and muscle fatigue resulting in increased ground reaction forces and loading rates may be associated with the high incidence of tibial stress fracture in the military. Interestingly, the percentage increase of the ground reaction loading rates were more pronounced than the increase of the ground reaction forces during the loaded and/or fatigued walking. With increased loading rates, there may be a reduction of the ground impact energy absorbed by leg muscles through their eccentric contractions. The lower extremity skeleton may have to absorb an increased portion of the ground impact energy. Thus, the risk of developing stress fracture may be increased. In fact, high vertical ground reaction loading rate was linked to tibial stress fractures in runners.⁶ Future research should address the possibility that it may be the increase in the mechanical loading rate rather than the load magnitude that contributes to an increased risk of tibial stress fractures.

A limitation of the study must be addressed here as the participants had no experience of walking with load carriage; the results from this study are only applicable to less-conditioned military recruits entering the Army basic training. It is possible that experienced load carriers may demonstrate different ground reaction force patterns that may not increase the likelihood of lower extremity injury during training. Future studies could focus on examining experienced load carrier's walking biomechanics and develop optimal training programs based on individuals experience and conditions.

In conclusion, under the influences of load carriage and muscle fatigue, participants demonstrated increased ground reaction forces and ground reaction loading rates during walking.

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Appendix J

Abstract presented at the 59th annual meeting of American College of Sports Medicine (ACSM) in San Francisco, CA. June 1- June 4, 2012.

The abstract was published in *Medicine & Science in Sports & Exercise*. V. 44, No. 5 Supplement, S282-283

Title: Influence of Physical Activity History on Ground Reaction Force during Walking

1788 Board #196 MAY 30 2:00 PM - 3:30 PM

Influence Of Physical Activity History On Ground Reaction Force During Walking

Henry Wang, Jeff Frame, Clark Dickin. *Ball State University, Muncie, IN.*

(No relationships reported)

Military recruits are commonly afflicted by lower-extremity overuse injuries such as tibial stress fracture (TSF). In particular, recruits with low levels of fitness are at a higher risk. Thus, it is advisable to precondition recruits before basic training. Running and basketball are common exercises to improve physical condition. However, patterns of ground impact loading are different between running and playing basketball. Multi-directional loading in basketball may lead to bone adaptation and reduce the risk of TSF. As loaded walking, a major task in basic training exposes recruits to high ground impact forces leading to increased risk of injury, it is important to examine if and how physical activity history (PAH) of running or basketball influences ground impact forces during loaded walking.

PURPOSE: To determine differences in vertical ground reaction force (VGRF) and loading rate (VLR) during loaded walking between runners and basketball players.

METHODS: Forty recreational runners ($n=20$, 21 ± 2 yr.) and basketball players ($n=20$, 21 ± 2 yr.) participated in this study. Participants completed four walking tasks in the following order: walking with 0kg (W00), 15kg (W15), 25kg (W25), and 35kg (W35) loads. Each task was performed for 5 min on a force instrumented treadmill (AMTI) at 1.67 m/s. Peak VGRF and VLR at weight acceptance were normalized to body weight (BW). Two-way repeated measures ANOVAs were performed. $\alpha = 0.05$.

RESULTS: No statistical differences in VGRF and VLR were found between the two groups ($P>0.05$). Increasing load carried had a significant effect on VGRF and VLR ($P<0.001$). As load carried increased, linear increases of VGRF (1.29 ± 0.06 , 1.57 ± 0.11 , 1.77 ± 0.22 , and 1.99 ± 0.19 BW for W00, W15, W25, and W35, respectively) and VLR (17.92 ± 3.72 , 22.51 ± 5.02 , 27.01 ± 8.33 , and 31.56 ± 7.18 BW/s for W00, W15, W25, and W35, respectively) were observed ($P<0.001$).

CONCLUSION: The significant increases of mechanical loading and loading rate are proportional to the increment of load carried. Despite differences in activity loading patterns between runners and basketball players, there were no differences in mechanical loading. Future studies should examine aspects of how PAH influences mechanical responses of tibia during loaded walking to improve the understanding of TSF. US ARMY#W81XWH-08-1-0587

Appendix K

Abstract presented at the 36th Annual Meeting of American Society of Biomechanics (ASB) in Gainesville, FL, August 14- August 18, 2012

Title: The Effects of Load Carriage and Fatigue on Frontal-plane Knee Mechanics during Walking

THE EFFECTS OF LOAD CARRIAGE AND FATIGUE ON FRONTAL-PLANE KNEE MECHANICS DURING WALKING

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INTRODUCTION

Military personnel are commonly afflicted by lower extremity overuse injuries [1, 2]. Overuse knee conditions are among the most common injuries during basic training [3]. Walking with heavy loads is an inevitable part of the military training, and during the twelve-weeks of basic training, the loaded running and walking distance could exceed 200 miles [1]. Therefore, military personnel have to face physical challenges comprised of load carriage and muscle fatigue.

During walking, the knee joint experiences an external adduction moment [4]. Large varus knee loading leads to cartilage degeneration and medial knee osteoarthritis (OA) [5,6,7]. Thus, the long-term effect of repetitive high varus knee loading could lead to medial knee OA; in the short term, walking with large varus knee loading could result in knee pain.

Load carriage increases vertical ground reaction force (GRF) during walking [8,9]. Walking in a fatigue state also results in increased vertical GRF [10]. It is possible that under the influences of load carriage and muscle fatigue, the knee joint may experience increased internal mechanical loading. However, it is unclear whether load carriage and fatigue result in an increase of varus knee loading during walking.

Analyzing frontal-plane knee mechanics during loaded and fatigued walking will broaden our knowledge on the potential causes of developing lower-extremity overuse injuries such as overuse knee conditions during military training.

The purpose of the study was to investigate the frontal-plane knee mechanics during loaded and fatigued walking. As the vertical GRF is increased during both the loaded and fatigued walking

[8,9,10], it was hypothesized that there would be increased internal knee abductor moments during loaded and fatigued walking.

METHODS

Eighteen healthy male subjects (age: 21 ± 2 yr.; body mass: 77.6 ± 9.6 kg; body height: 181 ± 4 cm) participated in the study. Subjects wore military boots and participated in a fatiguing protocol which involved a series of metered step-ups and heel raises while wearing a 16 kg rucksack. Subjects performed the following tasks in sequence: 5-min unloaded walking; 5-min loaded walking with a 32 kg rucksack; Fatiguing protocol; 5-min loaded walking with a 32 kg rucksack under fatigue; 5-min unloaded walking under fatigue. All walking tasks were performed at 1.67 m/s on a force instrumented treadmill (AMTI). A 15-camera system (VICON) was used to track reflective markers placed on the human body at 120 Hz. Ground reaction forces were collected at 2400 Hz. Visual 3D (C-Motion) was used to calculate lower extremity joint mechanics. The following variables were analyzed: peak hip and knee adduction angles, peak hip and knee abductor moments during weight acceptance of walking. Two-way repeated measures ANOVAs were performed. Load carriage and fatigue were the independent factors. $\alpha = 0.05$.

RESULTS AND DISCUSSION

No interactions were found between load carriage and fatigue for all the dependent variables ($P > 0.05$). Load carriage led to significant increases of hip adduction ($P < 0.05$), hip and knee abductor moments ($P < 0.001$) (Table 1). Fatigue did not lead to changes in hip and knee adduction angles and abductor moments ($P > 0.05$) (Table 1).

Frontal-plane knee mechanics is altered during loaded walking. There is a large internal abductor

moment introduced at weight acceptance. The increased internal abductor moment may be related to the increased GRF passing through medial side of the knee. As the internal abductor moment increases, medial compartment of the knee is under large compression. Increased stress in medial knee results in cartilage degeneration and onset of medial knee OA [5,6,7]. During a 12-week military training, the accumulated loaded walking/running distance exceeds 200 miles [1], the repetitive large medial knee loading may inflict cartilage damage in medial knee and result in knee pain.

In this study, we also found that the load carriage results in alterations of frontal-plane hip mechanics. There is an increase of hip adduction during weight acceptance of loaded walking. Increasing hip adduction stretches gluteus medius and enhances the muscle's ability to stabilize the pelvis. Indeed, large hip abductor moment is associated with loaded walking. However, increasing hip adduction also stretches tensor fasciae latae on the lateral side of the hip. As load carriage results in increased knee flexion at weight acceptance of walking [8], the friction between the lateral femoral condyle and the ilio-tibial band (ITB) could be elevated. Thus, during loaded distance walking, it is possible that increased hip adduction combined with cyclic knee flexion may lead to ITB syndrome.

Interestingly, the effect of fatigue on frontal-plane knee mechanics is insignificant. Although it was reported that there is an increased GRF associated with fatigued walking [10], the increased GRF may

be positioned close to the center of the knee. Thus, there is no alteration of the external adduction moment.

In summary, at weight acceptance, load carriage leads to alterations of frontal-plane hip and knee mechanics. The increases of hip adduction and knee abductor moment could be the causes of overuse knee conditions, which are common during military training.

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Table 1: Means and SDs of Peak hip and knee adduction angles and abductor moments during weight acceptance of walking.

Variables	Unloaded and Unfatigued	Loaded and Unfatigued	Loaded and Fatigued	Unloaded and Fatigued
Hip adduction angle (deg)*	9.3 (3.2)	10.3 (3.6)	10.2 (3.1)	9.5 (3.2)
Knee adduction angle (deg)	-1.4 (2.9)	-1.1 (2.9)	-1.4 (2.6)	-1.9 (2.9)
Hip abductor moment (Nm/kg)*	1.60 (0.18)	2.18 (0.38)	2.22 (0.35)	1.66 (0.24)
Knee abductor moment (Nm/kg)*	0.84 (0.18)	1.15 (0.32)	1.10 (0.36)	0.83 (0.22)

Note. * indicates significant difference between loaded and unloaded walking conditions ($P < 0.05$).

Appendix L

Abstract presented at the 60th annual meeting of American College of Sports Medicine (ACSM) in Indianapolis, IN. May 28 - June 4, 2013.

The abstract was published in *Medicine & Science in Sports & Exercise*. V. 45, No. 5 Supplement, S502

Title: The Influence of Physical Activity History on Ground Reaction Force during Running

2519 Board #220 May 31, 2:00 PM - 3:30 PM

The Influence of Physical Activity History on Ground Reaction Force during Running

Henry Wang, Clark Dickin, Ryan Hubble, Jeff Frame. *Ball State University, Muncie, IN.*
(No relationships reported)

Military recruits are commonly afflicted by overuse injuries such as tibial stress fracture (TSF). Low fitness levels is a risk factor for TSF, thus, it is advisable to precondition recruits before basic training (BT). Running and basketball are common forms of exercise for physical conditioning. The multi-directional loadings in basketball, vs. the unidirectional loadings in running may promote a different overall development of bone density and consequently different levels of the tibia's ability to resist unaccustomed loadings during BT. To date, the effect of physical activity history (PAH) of running or basketball on impact loadings is unclear. It is yet to be determined whether habitual runners adapt to running and exhibit lowered impact loadings, which may limit bone density gains overtime.

PURPOSE: To determine differences in the vertical ground reaction force (VGRF) and loading rate (VLR) during running between runners and basketball players.

METHODS: Forty recreational runners ($n=20$, 21 ± 2 yr.) and basketball players ($n=20$, 21 ± 2 yr.) participated in this study. Participants ran for 5 minutes on a force instrumented treadmill at 3.4 m/s. The VGRFs and VLR at stance were normalized to body weight (BW). One-way ANOVAs were performed. $\alpha = 0.05$.

RESULTS: Significant differences in VGRF and VLR were found between groups ($P<0.05$). The runners exhibited a lower impact VGRF (1.65 ± 0.05 BW vs. 1.81 ± 0.05 BW), active VGRF (2.43 ± 0.20 BW vs. 2.57 ± 0.16 BW), and VLR (81.62 ± 22.11 BW/s vs. 95.91 ± 16.12 BW/s) than those of the ball players.

CONCLUSION: Habitual runners can adapt to the running environment with decreased impact forces, which could lower risks of overuse injuries. This improved efficiency however, may result in reduced bone density levels in some areas. Thus, when a novel loading environment (e.g. loaded walking) is introduced, runners' tibia bones may need to accelerate the remodeling process to account for the altered loading. This may result in an increase of stress related bone problems (e.g. TSF) during BT. Accustomed to the multi-directional loading environment, ball players may be more resilient to uncustomary loadings during BT and have fewer TSF than runners. Future studies should examine the influence of PAH on tibia bone strength and density. US ARMY#W81XWH-08-1-0587

Appendix M

The following manuscript was published in Research Quarterly for Exercise and Sport, 84(3), 2013

Title: The Effects of Load Carriage and Muscle Fatigue on Lower-extremity Joint Mechanics

The Effects of Load Carriage and Muscle Fatigue on Lower-Extremity Joint Mechanics

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Daniel Leib and Eric L. Dugan
Boise State University

Military personnel are commonly afflicted by lower-extremity overuse injuries. Load carriage and muscular fatigue are major stressors during military basic training. **Purpose:** To examine effects of load carriage and muscular fatigue on lower-extremity joint mechanics during walking. **Method:** Eighteen men performed the following tasks: unloaded walking, walking with a 32-kg load, fatigued walking with a 32-kg load, and fatigued walking. After the second walking task, muscle fatigue was elicited through a fatiguing protocol consisting of metered step-ups and heel raises with a 16-kg load. Each walking task was performed at $1.67 \text{ m}\cdot\text{s}^{-1}$ for 5 min. Walking movement was tracked by a VICON motion capture system at 120Hz. Ground reaction forces were collected by a tandem force instrumented treadmill (AMTI) at 2,400 Hz. Lower-extremity joint mechanics were calculated in Visual 3D. **Results:** There was no interaction between load carriage and fatigue on lower-extremity joint mechanics ($p > .05$). Both load carriage and fatigue led to pronounced alterations of lower-extremity joint mechanics ($p < .05$). Load carriage resulted in increases of pelvis anterior tilt, hip and knee flexion at heel contact, and increases of hip, knee, and ankle joint moments and powers during weight acceptance. Muscle fatigue led to decreases of ankle dorsiflexion at heel contact, dorsiflexor moment, and joint power at weight acceptance. In addition, muscle fatigue increased demand for hip extensor moment and power at weight acceptance. **Conclusion:** Statistically significant changes in lower-extremity joint mechanics during loaded and fatigued walking may expose military personnel to increased risk for overuse injuries.

Keywords: gait, joint loading, kinetics

Military personnel are commonly afflicted by lower-extremity overuse injuries (Jones et al., 1993; Jones, Harris, Vinh, & Rubin, 1989; Knapik, Reynolds, & Harman, 2004). The high incidence of lower-extremity overuse injuries in the military is associated with high volume and intensity of physical training, during which the combined mileage of loaded walking and running often exceeds 200 miles (320 km; Jones et al., 1989). As large-magnitude ground reaction forces and loading rates have been related to lower-extremity injuries (Grimston, Engsberg, & Hanley, 1991; Jones et al., 1989; Milner, Ferber,

Pollard, Hamill, & Davis, 2006), large volumes of repetitive high ground-impact forces encountered during basic training could further increase the risk for lower-extremity overuse injuries in the military.

Common overuse injuries documented during basic training include knee pain, back pain, and stress fracture (Knapik et al., 2004). Severe injuries such as stress fractures demand extended periods of recovery and high medical costs (Brukner, Bennell, & Matheson, 1999). Although on the surface it seems that abnormal external loadings such as repetitive high-impact forces may contribute to lower-extremity overuse injuries, injury mechanisms are still far from being understood. In particular, the influences of training-related stressors such as load carriage and fatigue on lower-extremity joint mechanical loadings need to be examined.

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During military basic training, the strenuous training protocol exposes military recruits to significant muscular fatigue (Blacker, Fallowfield, Bilzon, & Willems, 2010). In addition, walking with a heavy load is always an important part of the training (Wilkinson, Rayson, & Bilzon, 2008). Prolonged walking with load carriage could lead to significant neuromuscular impairment (Blacker et al., 2010). Thus, muscular fatigue and load carriage are two major stressors experienced by military recruits. Although large external ground-impact forces associated with load carriage and muscle fatigue are linked to an increased risk for overuse injuries (Wang, Frame, Ozimek, Leib, & Dugan, 2012), alterations of lower-extremity joint mechanics could result in developing overuse injuries. Therefore, it is necessary to examine effects of fatigue and load carriage on lower-extremity joint mechanics during walking.

The effect of load carriage on gait mechanics has been investigated. It was reported that load carriage results in alterations of gait kinematics (Birrell & Haslam, 2009; Kinoshita, 1985). Carrying additional loads leads to reduced stride length, increased cadence, and increased pelvic tilt and hip flexion (Birrell & Haslam, 2009; Kinoshita, 1985). The effect of load carriage on gait kinetics was also analyzed with ground reaction forces as the focus (Birell, Hooper, & Haslam, 2007; Haman, Han, Frykman, & Pandorf, 2000; Kinoshita, 1985; Polcyn et al., 2002; Tilbury-Davis & Hooper, 1999; Wang et al., 2012). It was reported that load carriage results in increased vertical ground reaction forces (Birrell et al., 2007; Haman et al., 2000; Polcyn et al., 2002; Wang et al., 2012). The increase in vertical ground reaction forces is proportional to the amount of load carried (Polcyn et al., 2002; Tilbury-Davis & Hooper, 1999). In addition, load carriage leads to large increases of ground reaction loading rates; and the percent increase of loading rates is more pronounced than the percent increase of ground reaction forces (Wang et al., 2012). However, information related to the effect of load carriage on lower-extremity joint kinetics is limited. It is yet to be determined whether lower extremities increase power absorption to attenuate the increased ground-impact forces during loaded walking.

Skeletal muscles play an important role in attenuation of external loading. During loading response of running and drop landing, leg muscles contract eccentrically to attenuate ground-impact forces (Simpson, Ciapponi, & Wang, 1999). Muscular fatigue decreases muscle force generation and the muscle's ability to attenuate ground-impact forces (Verbitsky, Mizrahi, Voloshin, Treiger, & Isakov, 1998; Voloshin, Mizrahi, Verbitsky, & Isakov, 1998). Alterations of lower-extremity joint kinematics were observed during impact-related activities performed in a fatigued state. It was reported that there are changes in running kinematics associated with muscle fatigue (Christina, White, & Gilchrist, 2001; Derrick, Dereu, & McLean, 2002; Mizrahi, Verbitsky, Isakov, & Daily, 2000). Specifically, the knee joint becomes stiff with less flexion during weight acceptance (Mizrahi et al., 2000); the

ankle joint shows less dorsiflexion at heel contact (Christina et al., 2001). Moreover, changes in ground reaction forces and loading rate were also observed during activities performed in a fatigued state (Christina et al., 2001; James, Dufek, & Bates, 1994; Wang et al., 2012). In particular, muscular fatigue results in increases of vertical ground reaction forces and ground reaction loading rates during landing (James et al., 1994), running (Christina et al., 2001), and walking (Wang et al., 2012). To date, there is very limited information with regard to the effect of fatigue on lower-extremity joint kinetics during walking. It is yet to be determined whether there are alterations of lower-extremity joint loading during weight acceptance of fatigued walking.

Further, the combined effect of load carriage and muscular fatigue on lower-extremity joint mechanics needs to be examined. It is not clear if there is an interaction between these two factors. Determining the combined effect of fatigue and load carriage on lower-extremity joint mechanics will broaden our understanding of the mechanisms of lower-extremity overuse injuries.

The first purpose of this study was to examine the influence of load carriage on lower-extremity joint mechanics during walking. We hypothesized that there would be alterations of lower-extremity joint mechanics during loaded walking. Specifically, we expected there would be changes in joint angle, moment, and power in the sagittal plane during loading response. The second purpose of this study was to determine the effect of muscular fatigue on lower-extremity joint mechanics during walking. We hypothesized that there would be alterations in joint mechanics during fatigued walking. In particular, we expected to see changes in joint angle, moment, and power in the sagittal plane during loading response.

METHOD

Eighteen healthy college male participants were recruited to the study. The means and standard deviations (*SDs*) of age, body mass, body height, and maximal oxygen consumption ($\dot{V}O_{2\max}$) of participants were 21 years (*SD* = 2), 77.6 kg (*SD* = 9.6), 181 cm (*SD* = 4), and 51.4 ml·kg⁻¹·min⁻¹ (*SD* = 5.3), respectively. Participants were recreationally active, classified as low risk for cardiovascular diseases according to the American College of Sports Medicine (ACSM) guidelines (ACSM's *guidelines*, 2008), and free from known musculoskeletal injury. In addition, the age, body mass, body height, and fitness level ($\dot{V}O_{2\max}$) of the participants were comparable to military recruits entering the basic training program (Sharp et al., 2002). Institutional review board approval was obtained prior to commencing the study. Participants signed an informed consent document before testing.

A tandem force instrumented treadmill (AMTI, Advanced Mechanical Technology, Inc., Watertown, MA)

with two force platforms installed under the tandem belts was used to control the walking speed at $1.67 \text{ m}\cdot\text{s}^{-1}$ while allowing ground reaction forces to be collected. Reflective markers were attached on both sides of the body in the following locations to track the walking movement: acromion, sternum, anterior superior iliac spine, posterior superior iliac spine, lateral knee, lateral ankle, heel, base of the fifth metatarsal, and base of the second toe. In addition, two cluster marker sets were attached on the thigh and shank, respectively. Fifteen VICON MX and F-20s series cameras were used to track the reflective markers in the space. VICON NEXUS (V 1.4.116; VICON, Denver, CO) was used to collect kinematic data at 120Hz and ground reaction forces at 2,400Hz.

Participants wore compression shorts, a compression shirt, and military boots (Altama Mil-Spec Desert 3 Layer boot) during the experiment. Participants walked at a self-selected pace on the force instrumented treadmill for 5 min to warm up. After the warm-up, participants' maximal vertical jump heights were assessed by using a Vertec (Sports Imports, LLC, Columbus, OH). Three attempts were made and the highest of the three was used to determine 80% of the jump height. The general experimental protocol consisted of tasks in the following order: (a) 5-min normal walking in an unloaded and unfatigued state; (b) 5-min walking in a loaded and unfatigued state with a 32-kg rucksack (MOLLE, Specialty Defense System, Dunmore, PA); (c) fatiguing protocol; (d) 5-min walking in a loaded and fatigued state with a 32-kg rucksack; and (e) 5-min walking in an unloaded and fatigued state. The 32-kg load carriage used in this study represented a typical approach/marching load experienced by military personnel (Harman et al., 1999). The fatiguing protocol was carried out after the completion of the loaded walking task and was immediately followed by the loaded and fatigued walking task. No rest time was given between the tasks. Ten trials were collected during each walking task. A trial was defined as a 7-s data collection.

The goal of the fatiguing protocol was to induce a significant level of muscular fatigue in the lower extremities, defined as a decline of muscle force output in response to voluntary effort (Bigland-Ritchie & Woods, 1984). Therefore, participants completed circuits that included loaded stepping and heel raises. The stepping protocol was based on a modified Queens College Step Test

procedure (McArdle, Katch, & Katch, 2007). Specifically, the participant performed the test while wearing a 16-kg rucksack (MOLLE, Specialty Defense System, Dunmore, PA), the step height was set at 16 inches (40.64 cm), and the participant stepped up and down at a rate of 24 cycles per minute. One cycle was defined as step-up with first leg, step-up with second leg, step-down with first leg, and step-down with second leg. The first leg was alternated to ensure even loads across both legs. The participant performed this stepping sequence until he could no longer match the cadence of the metronome. At this time, the participant completed 20 heel raises standing at the edge of a box.

Following the heel raises, the participant removed the rucksack and completed a maximal-effort vertical jump using the Vertec (Sports Imports, LLC, Columbus, OH) to quantify the level of fatigue. This sequence was repeated until the participant's maximal vertical jump fell below 80% of his maximal jump height. Researchers provided verbal encouragement throughout the protocol to elicit maximal effort from the participant.

Experimental data were processed in Visual 3D Version 4.0 (C-Motion, Germantown, MD). Ground reaction forces were filtered using a zero-lag Butterworth filter with a cutoff frequency of 40 Hz. Kinematic variables analyzed were pelvis anteriotilt, hip flexion, knee flexion, ankle dorsiflexion at heel strike, and maximum knee flexion at weight acceptance. Kinetic variables included hip and knee extensor moments, ankle dorsiflexor moment, hip-joint power production, and knee and ankle-joint power absorption at weight acceptance. Joint moments and powers were normalized to body mass.

The Statistical Package for the Social Sciences Version 16 (SPSS, Inc., Chicago, IL) was used to perform statistical analysis. A two-way repeated-measures analysis of variance (ANOVA) was conducted to determine the effects of fatigue and load carriage. The sphericity assumption of the repeated-measures ANOVA is met as there are only two levels associated with the independent factors. Results were presented as means (SD). A-priori α was set at $< .05$.

RESULTS

Means and SDs of lower-extremity kinematics and kinetics during stance of walking are presented in Table 1 and

TABLE 1
Means and Standard Deviations (SDs) of the Lower-Extremity Joint Angles of the Unloaded and Unfatigued (UU), Unloaded and Fatigued (UF), Loaded and Unfatigued (LU), and Loaded and Fatigued (LF) Conditions During Weight Acceptance of Walking

	<i>UU</i>	<i>UF</i>	<i>LU</i>	<i>LF</i>
Pelvis anterior tilt at heel contact (degrees)	8.8 (5.9)	11.0 (8.6)	20.1 (5.0)	22.9 (8.6)
Hip flexion at heel contact (degrees)	32.1 (4.3)	28.2 (10.4)	45.4 (5.2)	40.6 (10.9)
Knee flexion at heel contact (degrees)	-2.5 (3.1)	-1.1 (4.5)	3.9 (3.2)	4.7 (4.9)
Maximum knee flexion at stance (degrees)	19.0 (2.8)	20.7 (4.4)	24.6 (4.5)	25.0 (5.3)
Ankle dorsiflexion at heel contact (degrees)	7.7 (1.9)	5.3 (4.9)	7.3 (2.7)	5.6 (3.6)

Table 2, respectively. Table 3 shows the *F* values, *p* values, and partial eta squared values of the lower-extremity joint angles during weight acceptance of walking. Table 4 shows the *F* values, *p* values, and partial eta squared values of the lower-extremity joint moments and powers during weight acceptance of walking.

There was no interaction between load carriage and muscle fatigue for all the dependent variables tested (Table 3 and Table 4). Load carriage had an effect on lower-extremity joint kinematics and kinetics. Kinematically, there were increases of pelvis anterior tilt, hip flexion, and knee flexion at heel contact and peak knee flexion at stance (Table 1). Kinetically, there were increases of hip and knee extensor moments, increases of hip-joint power production, and increases of knee and ankle joint power absorption (Table 2). Muscle fatigue also had an effect on lower-extremity joint mechanics. Kinematically, there was a decrease of ankle dorsiflexion at heel contact (Table 1). Kinetically, there were increases of hip extensor moment and joint power production and decreases of ankle dorsiflexor moment and joint power absorption (Table 2).

DISCUSSION

The purpose of the study was to assess effects of muscular fatigue and load carriage on lower-extremity joint mechanics during walking. In this study, muscular fatigue was introduced through a fatiguing protocol consisting of a series of stepping exercises and heel-raise exercises. During the fatiguing exercise, participants reached a fatigued state if they were not able to jump to 80% of their maximal pre-fatigue jump heights. A 32-kg rucksack was added to the body to represent the load carried. Participants performed the following four walking tasks: unloaded and unfatigued walking, loaded and unfatigued walking, loaded and fatigued walking, and unloaded and fatigued walking. The influences of muscular fatigue and load carriage on lower-extremity joint mechanics were then assessed.

In this study, we hypothesized that carrying an additional load would alter lower-extremity joint mechanics. As we expected, participants demonstrated increased pelvis tilt, knee and hip flexion at heel strike, and more knee flexion during weight acceptance. These findings are in agreement with previous research (Harman et al., 2000; Kinoshita,

TABLE 2
Means and Standard Deviations (SDs) of the Sagittal-Plane Lower-Extremity Joint Moments and Powers of the Unloaded and Unfatigued (UU), Unloaded and Fatigued (UF), Loaded and Unfatigued (LU), and Loaded and Fatigued (LF) Conditions During Weight Acceptance of Walking

	UU	UF	LU	LF
Hip extensor moment (Nm/kg)	1.54 (0.41)	1.85 (0.48)	2.26 (0.42)	2.38 (0.42)
Knee extensor moment (Nm/kg)	0.88 (0.20)	0.90 (0.25)	1.61 (0.37)	1.63 (0.42)
Ankle dorsiflexion moment (Nm/kg)	-0.44 (0.12)	-0.37 (0.10)	-0.45 (0.11)	-0.43 (0.11)
Hip-joint power production (W/kg)	1.03 (0.36)	1.47 (0.49)	1.69 (0.61)	1.97 (0.64)
Knee-joint power absorption (W/kg)	-1.34 (0.37)	-1.39 (0.54)	-2.65 (1.40)	-2.76 (1.06)
Ankle-joint power absorption (W/kg)	-1.04 (0.27)	-0.82 (0.32)	-1.12 (0.30)	-1.01 (0.35)

TABLE 3
F Values, *p* Values, and Partial Eta Squared Values of Two-Way Repeated-Measures ANOVA Tests for Sagittal-Plane Lower-Extremity Joint Angles During Weight Acceptance of Walking

Variables	Effects	<i>F</i> Value	<i>p</i> Value	Partial Eta Squared
Pelvis anterior tilt at heel contact	Load Carriage	70.938	.001	.807
	Fatigue	2.340	.145	.121
	Interaction	0.178	.678	.010
Hip flexion at heel contact	Load Carriage	289.700	.001	.945
	Fatigue	2.799	.113	.141
	Interaction	0.752	.398	.042
Knee flexion at heel contact	Load Carriage	45.563	.001	.728
	Fatigue	2.185	.158	.114
	Interaction	0.549	.469	.031
Maximum knee flexion at stance	Load Carriage	42.689	.001	.715
	Fatigue	3.294	.087	.162
	Interaction	3.078	.097	.153
Ankle dorsiflexion at heel contact	Load Carriage	0.002	.962	.000
	Fatigue	5.738	.028	.252
	Interaction	0.780	.389	.044

TABLE 4
F Values, *p* Values, and Partial Eta Squared Values of Two-Way Repeated-Measures ANOVA Tests for Sagittal-Plane Lower-Extremity Joint Moment and Power During Weight Acceptance of Walking

Variables	Effects	F Value	<i>p</i> Value	Partial Eta Squared
Hip extensor moment	Load Carriage	132.823	.001	.877
	Fatigue	1.521	.001	.477
	Interaction	4.203	.056	.198
Knee extensor moment	Load Carriage	95.382	.001	.849
	Fatigue	0.343	.566	.020
	Interaction	0.003	.960	.000
Ankle dorsiflexor moment	Load Carriage	3.469	.080	.169
	Fatigue	8.784	.009	.341
	Interaction	3.816	.067	.183
Hip-joint power production	Load Carriage	28.245	.001	.624
	Fatigue	16.190	.001	.488
	Interaction	1.250	.279	.069
Knee-joint power absorption	Load Carriage	28.600	.001	.627
	Fatigue	0.678	.422	.038
	Interaction	0.075	.787	.004
Ankle-joint power absorption	Load Carriage	7.381	.015	.303
	Fatigue	9.395	.007	.356
	Interaction	1.797	.198	.096

1985). The increased pelvis anterior tilt observed in this study could be a result of the increased trunk forward lean (Harman et al., 2000; Polcyn et al., 2002). Increasing trunk forward lean and pelvis anterior tilt may serve to vertically align the load carried with the body's center of mass (CM). Thus, participants can walk with improved sagittal-plane stability and make smooth weight transfers between legs.

The effect of load carriage on lower-extremity joint angles at heel strike and during weight acceptance of walking has not always been straightforward in the literature. The majority of load-carriage studies have shown increased knee and hip flexion at heel strike (Harman et al., 2000; Kinoshita, 1985). However, Tilbury-Davis and Hooper (1999) reported that sagittal-plane joint kinematics is not influenced by load carriage. Our study confirms that load carriage results in increased knee and hip flexion at heel strike and more knee flexion during weight acceptance. Increasing hip and knee flexion lowers the CM of the human body so that stability can be improved during walking (Harman et al., 2000). In addition, increasing knee flexion lengthens the quadriceps moment arm for greater production of knee extensor moment (Pandy & Shelburne, 1998). Thus, increasing knee flexion is a compensatory mechanism to enhance absorption of ground-impact forces. The inconsistent findings between Tilbury-Davis and Hooper's study (1999) and others may be due to participants' experience in load carriage. In Tilbury-Davis and Hooper's study, the recruited military personnel were experienced load carriers. In the current study and others (Harman et al., 2000; Kinoshita, 1985), participants had less experience or no experience with carrying heavy backpack loads. It is possible that experienced load carriers may be able to carry heavy loads with minimal kinematic changes

from their natural walking pattern and possibly lower the risk for injury. Finally, in this study, it was found that load carriage did not influence the ankle angle at heel strike. This is in agreement with the literature (Birrell & Haslam, 2009; Harman et al., 2000).

We also found that load carriage had a significant effect on lower-extremity joint moment. Both knee extensor and hip extensor moments were increased during weight acceptance. Also, there was a trend of increased ankle dorsiflexor moment ($p = .08$). Our findings are in agreement with Harman et al.'s study (2000), which reported that load carriage leads to increases of the lower-extremity joint moments. In this study, the increases of hip and knee extensor moments may correspond to the increased hip and knee flexion at heel strike. At the hip-joint level, increasing hip flexion prestretches the hip extensors and leads to an increase of hip extensor moment. At the knee-joint level, increasing knee flexion at initial contact not only serves to prestretch the knee extensors, but also lengthens the quadriceps moment arm (Pandy & Shelburne, 1998). Therefore, the production of knee extensor moment is enhanced.

We also found that load carriage had a significant effect on lower-extremity joint power. At the hip-joint level, there was an increase of power production at weight acceptance. The increased hip-joint power reflects an effort to return the anterior-tilted pelvis to its neutral position during stance. At the knee- and ankle-joint levels, greater power absorption was observed. As both the knee and ankle joints are responsible for shock absorption during weight acceptance (Rose & Gamble, 2006), it appears that the lower extremity increases power absorption to attenuate ground-impact forces at weight acceptance of loaded walking. However,

increased power absorption at the knee and ankle joints may elevate mechanical stresses in the knee extensors and ankle dorsiflexors.

Leg muscles play an important role in attenuating ground reaction forces (Simpson et al., 1999). When leg muscles are fatigued, their ability to dissipate ground reaction forces is reduced (Verbitsky et al., 1998; Voloshin et al., 1998). We had reported that leg muscle fatigue leads to increases of vertical ground reaction forces and loading rates during walking (Wang et al., 2012). From a mechanical point of view, changes in lower-extremity joint mechanics necessarily induce changes in ground reaction forces. Thus, in this study, we expected that there would be changes in lower-extremity joint mechanics during walking in a fatigued state. Indeed, we observed pronounced alterations of lower-extremity joint mechanics. Specifically, at the ankle-joint level, there were decreases of dorsiflexion at heel strike, dorsiflexor moment, and joint power absorption during weight acceptance; at the hip-joint level, there were increases of extensor moment and joint power production at weight acceptance.

The ankle joint plays an important role in facilitating human walking (Rose & Gamble, 2006). At weight acceptance, the ground reaction force vector passes behind the center of the ankle joint and creates an external plantarflexion moment. The ankle dorsiflexors contract eccentrically to counteract this external plantar-flexion moment and prevent the foot from slapping on the ground. The current study showed that under the influence of muscular fatigue, the dorsiflexors' ability to control the ankle-joint motion and absorb ground-impact forces was compromised. On one hand, there are increases of vertical ground-impact forces and loading rates (Wang et al., 2012). On the other hand, there is reduced ankle-joint function with decreases of dorsiflexion, dorsiflexor moment, and joint power absorption. Thus, large-impact loading may be transmitted to knee-joint level.

The knee joint was found to be the dominant joint used for shock attenuation in running (Derrick, 2004; Mizrahi et al., 2000). Increased knee flexion at heel strike of running is associated with increased shock attenuation (Derrick et al., 2002). In the current study, there was no increase in knee flexion at weight acceptance. The knee extensor moment and joint power absorption also stayed unchanged. Given the fact that there are significant increases of vertical ground-impact forces and loading rates during fatigued walking (Wang et al., 2012), lacking an adjustment in knee mechanics reflects the inability of the fatigued knee extensors to increase impact attenuation. Without increased contribution of shock attenuation from the knee extensors, the increased ground-impact forces may be further propagated to the upper body through the skeletal system. In fact, it has been suggested that increased shock propagation and decreased shock attenuation along the musculoskeletal system may lead to progression of overuse injuries such as osteoarthritis and low back pain (Collins & Whittle, 1989).

The hip extensors play a role to stabilize the pelvis and lift the CM of the body during weight acceptance of walking (Rose & Gamble, 2006). The hip extensor moment counteracts the external hip flexion moment introduced by the ground reaction force (Rose & Gamble, 2006). In this study, we found that both hip extensor moment and joint power production were increased during fatigued walking. As muscle fatigue induces increases of ground reaction forces (Wang et al., 2012), it appears that the increased demand for hip extensor moment and power is a result of the increased external hip flexion moment from the increased ground reaction force. Consequently, the hip extensors may experience increased mechanical stresses as well as an increased risk for muscle strain during fatigued walking.

The combined effect of load carriage and muscle fatigue on lower-extremity joint mechanics has not previously been examined. In this study, we found that there was no interaction between load carriage and muscle fatigue on lower-extremity joint mechanics. The combined effect of load carriage and muscle fatigue resulted in pronounced changes in joint kinematics and kinetics. Specifically, there were increases of pelvis tilt, knee, and hip flexion and a decrease of ankle dorsiflexion at heel contact. Also, both knee and hip joints experienced increases of extensor moment and power. It is possible that on one hand, carrying additional loads increases demand for large lower-extremity joint moments and powers; on the other hand, fatigued leg muscles experience difficulty in increasing force output (Bigland-Ritchie & Woods, 1984) to elevate joint moment and power. Thus, load carriage coupled with muscle fatigue could impose large mechanical stresses on the lower-extremity musculoskeletal system. The risk for developing lower-extremity overuse injuries would then be increased.

Stress fractures and knee pain are typical overuse injuries reported during military training (Jones et al., 1993; Knapik et al., 2004). In this study, load carriage was found to increase lower-extremity joint moments. An increase in joint moment reflects an increase of mechanical loadings in the joint (Andriacchi, Johnson, Hurwitz, & Natarajan, 2005), which may inflict joint conditions such as knee pain. In addition, bones under large mechanical loadings (e.g., compression and bending) would experience large bone strains (Nordin & Frankel, 2012). Thus, it is possible that during walking with load carriage, the repetitive mechanical loadings applied to the lower extremity would increase the risk for stress fractures in the tibia and femur. Muscle fatigue was found to reduce ankle joint power at weight acceptance. Reducing power absorption at the ankle implies that an increased portion of the kinetic energy at impact must be absorbed by lower-extremity bones and transmitted to proximal joints. Ground reaction loading rates have been shown to increase during fatigued walking (Wang et al., 2012). Thus, along with a possible increase in energy absorption, lower-extremity bones (e.g., tibia) could also experience fast rates of mechanical loadings and increased

strain rates. The risk for developing bone micro-damage is then increased. Therefore, the combined effect of load carriage and muscle fatigue could impose great mechanical stresses on lower-extremity bones. Repetitively applying such large mechanical loads at a fast rate to the bones could elicit tibial and/or femoral stress fractures. In fact, Milgrom et al. (2007) had speculated that the tibial bone strain would experience larger changes during fatigued running or marching when carrying heavy packs, which may result in an increased risk for tibial stress fracture. Future studies should examine deformation of lower-extremity bones under mechanical stresses from loaded and fatigued walking.

In summary, load carriage and muscular fatigue significantly influence lower-extremity joint mechanics during walking. Load carriage results in increases of pelvis anterior tilt, hip and knee flexion at heel contact, and an increased demand for lower-extremity joint moments and powers at weight acceptance. Muscle fatigue also leads to alterations of joint mechanics, which include decreases of ankle dorsiflexion, dorsiflexor moment and joint power, and increases of hip extensor moment and joint power at weight acceptance. The significant alterations of lower-extremity joint mechanics associated with load carriage and fatigue may help explain the high incidence of lower-extremity overuse injuries reported during army basic training.

As it is evident that load carriage and muscle fatigue—the two physical stressors experienced by military personnel during physical training—could elicit changes in lower-extremity joint mechanics, which may increase the risk for overuse injuries, it is necessary to explore ways to lessen the impacts of these two factors on military recruits entering basic training. The following recommendations may be considered. First, it seems logical to improve military recruits' leg muscle strength and endurance to delay the effect of muscle fatigue. Before entering basic training, based on outcomes of functional lower-extremity tests and/or military screening tests, introducing a custom weight-training program consisting of resistance and endurance trainings will be helpful to the recruits. Second, it is evident that experienced load carriers are able to minimize changes in joint kinematics (Tilbury-Davis & Hooper, 1999), which may be associated with smaller increases of joint loading and lower risks for overuse injuries. It is reasonable to encourage recruits to practice load carriage before entering the training to become familiar with loaded walking. Also, gradually increasing the load carried through the course of basic training may be desirable.

Some limitations of the study must be addressed here. First, although the fatiguing protocol used in this study effectively elicited muscle fatigue in the participants, it is different from situations in basic training, during which recruits develop fatigue over a prolonged period of time (e.g., hours and days). It is possible that recruits may exhibit greater alterations in joint mechanics during the actual

training than during the laboratory testing. Second, participants in this study had no experience with load carrying. Results from this study are more applicable to recruits entering basic training with limited experiences of load carriage.

WHAT DOES THIS ARTICLE ADD?

Although epidemiologic studies have revealed that lower-extremity overuse injuries are common among military recruits, the injury mechanism is far from being understood. This study examined effects of load carriage and muscle fatigue on gait mechanics and determined that there are significant alterations of lower-extremity joint kinematics and kinetics. Significant changes in joint mechanics could result in increased mechanical stresses imposed on the musculoskeletal system. Thus, the high incidence of lower-extremity overuse injuries seen in military recruits can be related to the pronounced alterations of joint mechanics during loaded and fatigued walking. Outcomes from this study could help develop exercise programs to precondition military recruits before entering basic training to reduce the incidence of overuse injuries.

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Appendix N

Abstract presented at the 36th Annual Meeting of American Society of Biomechanics (ASB) in Omaha,
NE, September 3 – September 5, 2013

Title: The Effects of the Type of Activity on Tibial Strain Characteristics

CHARACTERISTICS OF TIBIAL STRAINS DURING DIFFERENT TYPE OF PHYSICAL ACTIVITIES

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INTRODUCTION

There is evidence that an individual's past physical activity influences their risk of sustaining a tibial stress fracture. This evidence is based primarily on epidemiologic research on the rates of injury in different sub-populations [1]. The mechanisms that may explain these results have not been adequately examined. The goal of this study was to develop muscle driven forward dynamics simulations to investigate the influence of lower extremity exercises on tibial strains. The study hypothesized that high impact activities such as drop jump and cutting maneuvers produce different tibial strain profiles than those produced during walking. Experimentally measured kinematic data and ground reaction forces were used as inputs to the simulations while the tibial strain values were extracted.

METHODS

One healthy male subject (Age = 19 yr., height = 1.80 m, and weight = 80 kg) performed the following four different type of exercises: drop-jump (JUM), cutting maneuver (CUT), running (RUN) and walking (WAK). A VICON motion capture system was used to record kinematics (240 Hz) and AMTI force plates were used to record ground reaction forces (2400 Hz). Computed tomography (CT) images of the subject were obtained in order to develop the right tibial bone geometry for a subject-specific lower extremity model. The 3D tibia was segmented in MIMICS 14.0 (Materialise, Leuven, Belgium).

MARC 2012 (MSC.Software, Santa Anna, CA) was used to develop a finite element (FE) model of the right tibia bone. Mechanical properties were assigned based on bone density. FE model was converted into a flexible tibia in MARC to incorporate the tibia geometry in LifeMod (LifeModeler Inc., San Clemente, CA). The subject's weight, height, gender, and age as well as the relative positions of the ankle, knee, and hip joints, determined from the motion

capture, were used to scale the generic lower extremity models based on the GeBOD anthropometric database. The generic right tibia bone geometry was replaced with the developed flexible body of the subject specific tibia bone geometry. Tri-axis hinges combined with passive torsional spring-dampers were employed to model the hip joints. A hinge joint with a single degree of freedom was used for knee and ankle joints in the sagittal plane. A total of ninety muscles were added to the right/left legs. The measured kinematics, collected during lower extremity exercises, was used to drive the model with an inverse dynamics algorithm [2] while the muscle shortening/lengthening patterns were recorded. Next, kinematic constraints were removed, and muscles served as actuators to replicate the motions during forward dynamics (Fig.1).

A proportional-integral-derivative (PID) feedback controller was implemented to calculate each muscle force magnitude using the error signal between the current muscle length in the forward dynamics and the recorded muscle length during the inverse dynamics simulation. The force generated by individual muscle was limited by its force generating potential given by the following equation:

$$F_{max} = PCSA \times \sigma_{max} \quad (1)$$

Where F_{max} is the muscle's maximum force, $PCSA$ is the physiological cross sectional area and σ_{max} is the maximum tissue stress.

The maximum and minimum principle strain values (tension/compression) for all surface nodes of the tibial bone were computed in ADAMS (MSC. Software, Santa Ana, CA) during simulations. In order to compare the tibial strains during different activities in a consistent way, only maximum (*Max*) and average (*Mean*) strain values were reported for the nodes within mid-medial tibial shaft region [3].

RESULTS AND DISCUSSION

Although developing more computational models are in the process, the presented results are based on only one single subject specific model. Figure 2 illustrates tension (Fig.2a) and compression (Fig.2b) strains in color coded bars respectively during the stance phase of the cutting maneuver exercise. The average model predictions over the three trials of cutting maneuver are shown in figure 3 with a solid line and a shaded area corresponding to ± 1 standard deviation. Table 1 summarizes the maximum and average values for the tibial strain on the mid-medial tibial shaft region over all four exercises during the stance phase.

Table 1: Tension (TEN) and Compression (COM) strain values during four different types of activities. COM strain values are rectified.

micro STRAIN	JUM	CUT	RUN	WAK	
TEN	Max	1265	923	950	454
	Mean	390	580	338	206
COM	Max	437	1257	334	504
	Mean	179	572	162	212

This study produced a subject specific musculoskeletal model capable of concurrent simulation of muscle driven forward dynamics and calculated the tibial strain values. The current study shows that different types of physical activities exhibit different tibial strain profiles. High impact activities such as drop-jumping, cutting, and running elicit large tibial strain. Thus, participating in high impact activities may introduce mechanical stimulations to bone and lead to positive bone adaptations to resist unaccustomed loading environments. In summary, this modeling technique can provide useful insights of bone reactions to mechanical loadings during dynamic activities.

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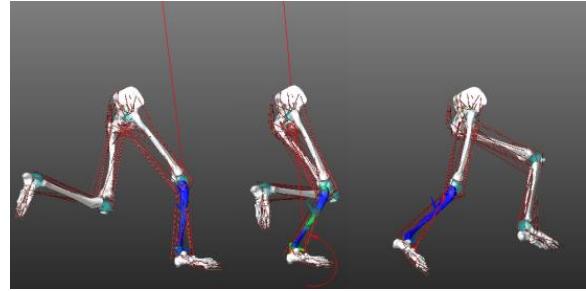


Figure 1: Lower extremity multi-body model of the subject with the flexible right tibial bone during the cutting maneuver.

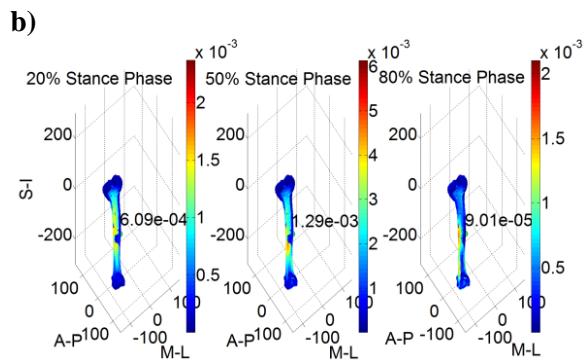
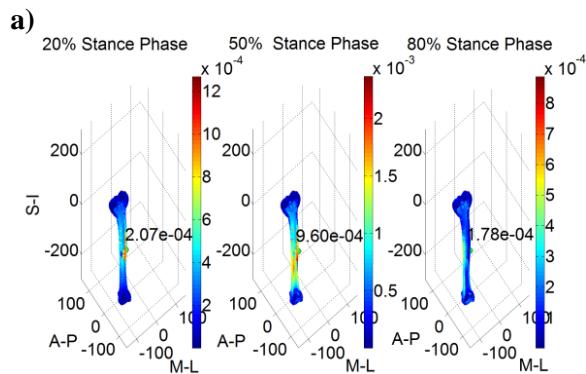


Figure 2: Computed strain at 20 %, 50% and 80% stance phase of the cutting maneuver.

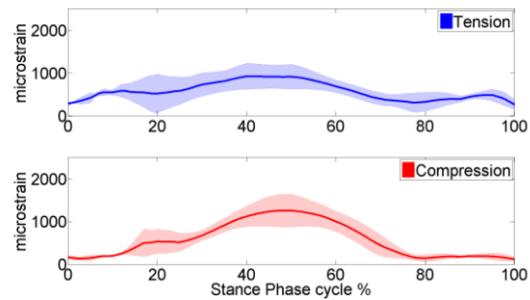


Figure 3: Model predicted strain on the mid-medial tibial shaft region.